Detecting kinematic gait abnormalities in people with multiple sclerosis using clinically practical measures

A Thesis Submitted to the

College of Graduate Studies and Research

in Partial Fulfillment of the Requirements

for the Degree of Master of Science

in the College of Kinesiology

University of Saskatchewan

Saskatoon

By

Kristopher Blaine Beyer

Permission to use

In presenting this thesis/dissertation in partial fulfillment of the requirements for a Postgraduate degree from the University of Saskatchewan, I agree that the Libraries of this University may make it freely available for inspection. I further agree that permission for copying of this thesis/dissertation in any manner, in whole or in part, for scholarly purposes may be granted by the professor or professors who supervised my thesis/dissertation work or, in their absence, by the Head of the Department or the Dean of the College in which my thesis work was done. It is understood that any copying or publication or use of this thesis/dissertation or parts thereof for financial gain shall not be allowed without my written permission. It is also understood that due recognition shall be given to me and to the University of Saskatchewan in any scholarly use which may be made of any material in my thesis/dissertation.

Requests for permission to copy or to make other uses of materials in this thesis/dissertation in whole or part should be addressed to:

Head of the Department of Kinesiology
University of Saskatchewan
Saskatoon, Saskatchewan
S7N 5B2

Abstract

The effects of multiple sclerosis (MS) on the central nervous system often manifest as abnormalities in gait kinematics. Clinically practical, valid, and reliable measures of gait kinematics are necessary to address research and clinical questions about MS. Wireless flexible electrogoniometry (EG) is a clinically practical measure of joint angles. The GAITRite walkway system is a clinically practical, valid and reliable measure of temporal and spatial gait characteristics. The overall objective of this two-study research project was to explore whether these clinically practical measures of gait kinematics can be used to accurately detect gait abnormalities in people with multiple sclerosis. Study 1 examined the reliability and validity of EG and Study 2 examined the gait kinematics of people with MS (PWMS) using EG and GAITRite. For Study 1, angle at initial contact and total joint excursion were measured by EG at both the knee and ankle while ten healthy adults walked at a self-selected comfortable speed. Measurements were repeated for two testers and two visits to assess reliability. The same variables were measured concurrently with three-dimensional motion analysis (3D) to assess validity. For all variables, reliability was good as indicated by low measurement error and validity was good as indicated by association and agreement of EG with 3D. For Study 2, the same joint angles, along with speed, cadence, step length, stride length, stance duration and double support duration were assessed for six PWMS and six controls without MS. PWMS showed significantly reduced speed, cadence, and ankle excursion and increased stance and double support duration as previously shown with 3D. Spasticity and/or instability may lead to these kinematic gait abnormalities in PWMS; however, reduced velocity may confound this interpretation by affecting the other observed gait abnormalities. Further research about the determinants of gait dysfunction in PWMS is required. EG and GAITRite are clinically practical, valid and reliable measures of gait kinematics and should be included in further clinic-based research to determine which kinematic gait abnormalities are causes and which are effects of the observed decrease in gait speed in PWMS.

Acknowledgements

When I started this project I was fortunate to have the love and various forms of support of my family, who showed unconditional faith in me without, at times, even a hint of what I was getting myself into. This is largely because I, myself, did not have a hint of what I was getting into. At the beginning, this project was an entirely unknown beast, like the elephant in the fable, making me one of the blind men charged with describing it.

Along the way, however, I was blessed to share the experience with fellow graduate students who were all too willing to provide their various opinions and ideas about research and to offer their support when it was needed. I would be remiss not to specifically acknowledge Mike Smith who also shared his time and knowledge by helping me to develop and implement my data collection protocol. I was, of course, also fortunate to receive financial support from the College of Kinesiology, the University of Saskatchewan, the Natural Sciences and Engineering Research Council, and CRC Training Funds.

I also need to thank and acknowledge the contributions of Dr. Katherine Knox, my committee members, Dr. Gord Binsted and Dr. Joel Lanovaz, and my external examiner, Dr. Gary Linassi, each of whom presented various parts of the figurative elephant allowing me to view it from many different perspectives. Their knowledge and patience is truly appreciated.

Finally, I owe a debt of gratitude to my supervisor, Dr. Larry Brawley, for nudging me toward a better understanding of the elusive elephant by, at appropriate times, integrating or separating the various perspectives presented to me. He also allowed me to develop as a researcher by encouraging the cultivation of my strengths and by subtly challenging my weaknesses. Without his guidance, I would still be blindly groping at an unknown beast.

Table of contents

List of appendices

List of figures

List of tables

1. General introduction

1.1. Walking and gait

Walking depends on the continuous interaction of sensory and motor information in the central nervous system (CNS; Grillner, Wallén, Saitoh, Kozlov, & Robertson, 2008; Rossignol, Dubuc, & Gossard, 2006). In humans (Borelli, 1680 as cited in Clarac, 2008; Borelli, 1989 as cited in Ashley-Ross & Gillis, 2002), as with other mammals (Graham Brown, 1911), stepping is caused by the rhythmic alternation of contractions in flexor and extensor muscles in the legs. In some mammals, this rhythmic alternation is produced and coordinated by a spinal central pattern generator (Ferdirchuk, Nielsen, Petersen, & Hultborn, 1998; Jankowska, Jukes, Lund, & Lundberg, 1967 as cited in Hultborn & Nielsen, 2007), a network of neurons within the spinal cord that controls the sequence of muscle activation for a specific motor pattern such as walking (Graham Brown, 1911). Similar networks are suspected to exist in humans but have not been explicitly located despite an abundance of circumstantial evidence (Hultborn & Nielsen, 2007). Descending signals, sent from locomotor regions in the cerebellum and brain stem via the reticulospinal pathway to the spinal cord, initiate the rhythmic walking pattern and control its speed (Shik, Severin, & Orlovsky, 1966 as cited in Rossignol et al., 2006). At rest, the locomotor regions of the brainstem are inhibited by the basal ganglia causing the central pattern generators to be silent (Garcia-Rill, 1986; Takakusaki, Habaguchi, Ohtinata-Sugimoto, Saitoh, & Sakamoto, 2003). Descending signals from the cerebral cortex to the basal ganglia can lift this inhibition to initiate walking (Takakusaki et al., 2003).

Sensory input modifies the motor signals to coordinate timing, adjust to obstacles, maintain balance, and steer (Grillner et al., 2008; Rossignol et al., 2006). Proprioceptors within the muscles and joints detect the rate of stretch of muscles and the amount of tension in tendons to provide information about the relative position of the limbs; exteroceptors in the skin detect touch, temperature, and pressure to provide information about the external environment (Sherrington, 1906). This somatosensory input is incorporated both directly at the spinal cord and via the cerebellum, which sends corrective signals to the locomotor regions of the brain stem, to coordinate the timing and amplitude of stepping and position the foot during normal gait and in the event of a disturbance (Rossignol et al., 2006). The vestibular complex in the inner ear detects linear and angular accelerations of the head (Sherington, 1906) and is incorporated directly into the reticulospinal pathway to coordinate posture (Grillner et al., 2008; Rossignol et

al., 2006). During locomotion, the visual system must assess constantly changing input described as an optic flow field, which provides information about the movement of the environment relative to the body, in order to adjust speed, maintain posture, steer, and adjust to objects (Lee, 1980). Visual input is incorporated via the brainstem or the motor cortex (Grillner et al., 2008; Rossignol et al., 2006). Gait, the pattern of walking resulting from the interaction of this motor and sensory information, may be affected by disturbances in the CNS.

1.2. Multiple sclerosis

Multiple sclerosis (MS) is a disease of the CNS (Charcot, 1868 as cited in McDonald, 1974) that affects the transmission of motor (Ng, Miller, Gelinas, & Kent-Braun, 2004) and sensory (Cameron, Horak, Herndon, & Bourdette, 2008) signals and results in walking difficulties (Paltamaa, Sarasoja, Wikström, & Mälkiä, 2006; Scheinberg et al., 1980). MS is characterized primarily by damage to the myelin of the nerves in the CNS (Charcot, 1868 as cited in McDonald, 1974). A sheath of myelin increases the speed of signal transmission along the axons of both sensory and motor neurons; therefore, loss of myelin results in slowed sensory and motor signal transmission (Schwartz & Westbrook, 2000). This demyelination may also be accompanied by damage to the axon itself (Charcot, 1868 as cited in McDonald, 1974; Ferguson, Matyszak, Esiri, & Perry, 1997) and the oligodendrocytes responsible for remyelinating the axon (Morales, Parisi, & Lucchinetti, 2006; Rinker, Naismith, & Cross, 2006). The damage is presumed to be caused by an inflammatory immune response to a myelin antigen as evidenced by the presence of T-cells and macrophages at the lesion site (Rinker et al., 2006). Studies of twins, families and races suggest that certain people may have a genetic susceptibility to MS but how this is passed on is still unclear (Pryse-Phillips & Sloka, 2006). Some but not all of the people with this genetic susceptibility may then develop MS lesions in response to an external attack or attacks which may be environmental, infectious or autoimmune in nature.

The clinical course of MS is generally classified into 4 subtypes (Lublin & Reingold, 1996): relapsing-remitting (RR), primary-progressive (PP), secondary-progressive (SP), and progressive-relapsing (PR). RR-MS is characterized by relapses, in which symptoms are either subjectively reported or objective measured to worsen for a period lasting longer than 24 hours (McDonald et al., 2001; Poser et al., 1983), followed by a varying amount of recovery and longer periods of remittance (Lublin & Reingold, 1996). Nearly 85% of patients are classified as RR-

MS at onset (Weinshenker et al., 1989). PP-MS is characterized by the gradual worsening of symptoms without relapses although brief periods of remittance may occur (Lublin & Reingold, 1996). Most patients classified as RR-MS at initial onset will eventually also convert to a course of gradual worsening similar to that in PP-MS (Weinshenker et al., 1989) defined as SP-MS (Lublin & Reingold, 1996). Finally, in rare instances, patients may be classified as PR-MS if they experience a gradual worsening of symptoms from onset accompanied by relapses with or without remittance (Lublin & Reingold, 1996).

People with MS may experience a diverse array of signs and symptoms depending on the amount and location of demyelination and axonal damage (Pryse-Phillips & Sloka, 2006). Disability in people with MS is usually quantified by the Expanded Disability Status Scale (EDSS). The EDSS is determined by the evaluation of neurological impairment in 9 different Functional Systems representing the range of signs and symptoms associated with MS (Kurtzke, 1983). These include, but are not limited to, the following: weakness and paralysis (Pyramidal); uncoordinated movement (Cerebellar); problems speaking or swallowing (Brain Stem); decreased sensation of touch, temperature or proprioception (Sensory); bowel and bladder dysfunction (Bowel and Bladder); vision problems (Visual); cognitive impairment or mood alteration (Cerebral; Kurtzke,1983).

1.3. Effects of multiple sclerosis on gait

People with multiple sclerosis (MS) have ranked walking as the most important physical function ahead of vision, lack of pain, and thinking and memory (Heesen et al., 2008). Alas, in one study MS patients ranked balance problems and walking difficulties as the second and third most common major symptoms affecting daily life, behind only fatigue (Paltamaa et al., 2006). In another study, gait and motor disturbances were the greatest complaints of 85% of MS patients interviewed at an outpatient clinic in New York (Scheinberg et al., 1980). This is disturbing since the loss of mobility has been shown to negatively influence the performance of activities of daily living and quality of life for people with MS (Sutliff, 2010).

MS lesions may affect gait at various sites within the CNS; their frequency in the optic nerves, brain stem, cerebellum, and cortical and spinal cord white matter (Noseworthy, Lucchineti, Rodriguez, & Weinshenker, 2000) means that the areas of the CNS responsible for gait are often targets for lesions. This is supported by the high frequency of involvement of the

Pyramidal (84.9%), Cerebellar (76.9%), Brain Stem (73.0%), and Sensory (55.2%) Functional Systems on the standard neurologic examination (Kurtzke, 1983). In fact, six of the eight functional systems tested by the standard neurological exam present the potential to negatively affect gait if involved.

The most likely mechanism through which MS affects gait is the transmission impairment of both ascending sensory and descending motor signals. People with MS have demonstrated slowed spinal somatosensory conduction, which has been associated with delays in postural responses to maintain balance (Cameron et al., 2008). Presumably, transmission of somatosensory information during walking may also be slowed reducing the ability of the CNS to coordinate the timing and amplitude of stepping and to avoid obstacles. People with MS have also shown weakness in voluntary maximal ankle dorsi-flexion, a slower rate of voluntary force development, and a reduced ability to perform rapid successive foot-taps compared to controls without MS (Ng et al., 2004). These characteristics are present despite similar electrically stimulated maximal contraction and muscle cross-sectional area indicating impairment of descending signal transmission within the CNS (Ng et al., 2004). This impairment could affect the ability of the CNS to initiate and maintain a rhythmic walking pattern or control its speed; it may also lead to spasticity (Morita, Crone, Christenhuis, Petersen, & Nielsen, 2001). Spasticity is an increase in muscle tone caused by hyperexcitability of the stretch reflex (Lance, 1980), which may be due to a reduced ability to modulate the excitability of the stretch reflex via descending pathways (Morita et al., 2001). In people with MS, spasticity has been associated both postural dysfunction (Sosnoff, Shin, & Motl, 2010) and walking disability (Rizzo, Hadjimichael, Preiningerova, & Vollmer, 2004).

1.4. Kinematic gait analysis

Kinematic gait analysis describes the linear and angular displacements, velocities, and accelerations of body segments while a person is walking and can serve a number of research and clinical purposes (Coutts, 1999; Winter, 2005). Often, kinematic gait abnormalities in clinical populations with movement disorders (e.g., Parkinson's disease: Chien et al., 2006; Nelson et al., 2002 or MS: Givon, Zeilig, & Achiron, 2009) are characterized and followed over time to examine basic research questions either about the disease or how it affects the various systems involved in walking (i.e., musculoskeletal, nervous, etc.). Clinically, kinematic gait

analysis is often part of a detailed patient assessment meant to diagnose and characterize specific movement disorders in order to initiate and evaluate appropriate treatment (Coutts, 1999; Krebs, Edelstein, & Fishman, 1985; Whittle, 1996). For a measure of kinematic gait analysis to be valuable clinically, it must not only be valid (i.e., it is measuring the true quantity it was designed to measure and is sufficiently free from random error) and reliable (i.e., it consistently measures this true quantity over repeated measures with minimal variance due to random error; Altman & Bland, 1983; Thomas, Nelson, & Silverman, 2005; Vincent, 2005) but also be clinically practical (i.e., it is easy to administer, acceptable to both patients and health care professionals, and resource efficient; Rudick et al., 1996).

Observational gait analysis is the most commonly used method of gait analysis among clinicians because it is simple and convenient, not requiring the use of expensive equipment (Coutts, 1999; Krebs et al., 1985; Whittle, 1996; Yack, 1984). When performed by an experienced clinician, observational gait analysis provides a good overall qualitative impression of a patient's gait (Coutts, 1999; Yack, 1984) that is useful in identifying gait abnormalities in most situations (Whittle, 1996). It is, however, a difficult task that requires the ability to observe movement in multiple planes during the rapidly repeating gait cycle (Saleh & Murdoch, 1985). Consequently, it has shown poor inter- and intra-rater reliability (Coutts, 1999; Krebs et al., 1985) and poor agreement with objectively measured quantitative gait parameters (Kawamura et al., 2007; Saleh & Murdoch, 1985; Williams, Morris, Schache, & McRory, 2009).

In the research laboratory, three-dimensional motion analysis (3D) is the current 'gold' standard in kinematic gait analysis (Coutts, 1999). It is both reliable (McGinley, Baker, Wolfe, & Morris, 2009) and valid (Benoit et al., 2006; Stagni, Fantozzi, Cappello, & Leardini, 2005) when calculating sagittal plane joint angles while walking. Evaluation of movement kinematics with 3D allows for the identification of movement and postural abnormalities earlier than other clinical examinations in people with MS (Benedetti et al., 1999; Corradini, Fioretti, Leo, & Piperno, 1997; Martin et al., 2006; Solaro et al., 2007). Despite its advantages, 3D requires resources such as space, time, money, and expertise that make it impractical for most clinicians (Krebs et al., 1985). As well, analysis of specific kinematic variables may narrow the focus of the observer, spurning the clinician's extensive experience with the disorder and their subjective assessment (Yack, 1984). This should not be the case; rather, objective measures of gait kinematics should be coupled with and validated by observations made by the clinician to

provide a rich description of the movement (Saleh & Murdoch, 1984; Yack, 1984). There is a clear need for objective measures of gait kinematics that are not only valid and reliable but also clinically practical.

1.5. Overall purpose of the current studies

Kinematic analysis of gait in people with MS may lead to knowledge about how the CNS controls gait and how MS affects the CNS. If a practical measure exists, it may also provide a valuable clinical tool to help make more informed diagnostic and treatment decisions. Accordingly, the overall objective of this two-study research project was to explore whether clinically practical measures of gait kinematics can be used to accurately detect gait abnormalities in people with multiple sclerosis. The next two sections describe two separate studies that explore this objective. The first study examines the reliability and validity of knee and ankle joint angle measurement by a clinically practical measure, flexible electrogoniometry, during walking. The second study evaluates the feasibility of using two clinically practical measures of gait kinematics, the GAITRite walkway system and flexible electrogoniometry, to detect gait abnormalities in people with MS.

2. Study 1: Reliability and validity of knee and ankle joint angle measurement by flexible electrogoniometry during walking

2.1. Introduction

Valid, reliable, and clinically practical objective measures exist for some but not all kinematic gait parameters. For example, the GAITRite walkway system is an objective measure that is quick, portable, relatively inexpensive, and simple when compared to other objective assessments such as 3D (van Uden & Besser, 2004). It has also demonstrated reliability (Bilney, Morris, & Webster, 2003; Menz, Latt, Tiedemann, Kwan, & Lord, 2004 ; van Uden & Besser, 2004) and concurrent validity with a number of established measures (Bilney et al., 2003; Cutlip, Mancinelli, Huber, & DiPasquale, 2000; McDonough, Batavia, Chen, Kwon, & Ziai, 2001; Webster, Wittwer, & Feller, 2005). The GAITRite system can effectively be used to measure temporal and spatial gait parameters and has been used to detect gait abnormalities in people with multiple sclerosis (Givon et al., 2009) and Parkinson's disease (Chien et al., 2006; Nelson et al., 2002).

Flexible electrogoniometry (EG) is a clinically practical objective measure of lower body joint angle; it is lightweight, portable, simple, and inexpensive (Moriguchi, Sato, & Coury, 2007; Piriyaprasarth, Morris, Winter, & Bialocerkowski, 2008; Rowe, Myles, Hillmann, & Hazlewood, 2001). However, compared to other clinically practical measures such as the GAITRite walkway, EG has limited evidence for its reliability and validity during gait. EG has been shown to be a reliable measure of knee joint angle in various static positions (Piriyaprasarth et al., 2008) and even at initial contact during level walking (van der Linden, Rowe, & Nutton, 2008). It has also shown concurrent validity with 3D when measuring sagittal joint angles at the knee while walking (Rowe et al., 2001) and at the ankle while performing common dance movements (Agraharasamakulam, Bronner, & Ojofeitimi, 2005). Further examination of the reliability of EG at the knee during walking is required and confirmation of its concurrent validity with 3D would be beneficial. EG has already been used to quantify ankle joint motion during treadmill walking in healthy adults (Moriguchi et al., 2007); however, EG currently lacks sufficient evidence of both its reliability and validity at the ankle during normal walking on a flat surface. Establishing the reliability and validity of a measure is essential in determining whether or not measurements made by it can be trusted (Thomas et al., 2005).

Any technique designed to indirectly measure a certain quantity will observe the true value of that quantity plus or minus some amount of random error (Thomas et al., 2005). As such, the variance in the observed quantity is also caused not only by variance in the true quantity but also in the amount of random error (Weir, 2005). The true quantity obviously cannot be known so, functionally, it may also contain an amount of predictable or anticipated error as a result of indirect measurement that cannot be avoided (Harvill, 1991). To determine whether a new measure can be used in place of or interchangeably with an established technique we must determine that the new technique is valid (i.e., it is measuring the true quantity it was designed to measure and is sufficiently free from random error) and reliable (i.e., it consistently measures this true quantity over repeated measures with minimal variance due to random error; Altman & Bland, 1983; Thomas et al., 2005; Vincent, 2005). A measure can be reliable without being valid but the level of reliability limits the level of validity possible (Bland & Altman, 1999).

Reliability can be expressed in relative (i.e., consistency of rank of score) or absolute (i.e., consistency of actual score) terms (Weir, 2005). Relative consistency is assessed using reliability coefficients, calculated as a ratio of true score variance to obtained score variance. Intraclass correlation (ICC; Shrout & Fleiss, 1979) is more appropriate than interclass correlation (i.e., Pearson *r*) for determination of relative consistency because ICC compares two (or more) repeated measures of the same variable and provides estimates of the different sources of variance (Thomas et al., 2005). Absolute consistency is assessed by measuring the precision of the scores, calculated as the variation in the measurement error or standard error of measurement (SEM). When assessing reliability it is important to assess and report both relative (i.e., ICC) and absolute (i.e., SEM) consistency (Harvill, 1991; McGinley et al., 2009; Weir, 2005). Absolute consistency is especially helpful for clinicians trying to differentiate real change from that due to error (Eliasziw, Young, Woodbury, & Fryday-Field, 1994).

When the true measurement cannot be known, the criterion validity of the new measurement technique must be determined by comparing it to an established technique (Thomas et al., 2005). When the new and established techniques are measured concurrently, we can examine their concurrent validity by seeing if they are in agreement on the measured quantity. While interclass correlations (e.g., Pearson *r*) are often used to assess the agreement of two measurement techniques, they are actually a measure of association, not agreement (Altman & Bland, 1983; Bland & Altman, 1986). Correlations assess whether there is a relationship

between the variance of the measurements by the two techniques; that is, whether quantities that are distant from the mean when measured by the established technique are also distant from the mean when measured with the new technique. This indicates the association of the two techniques but tells us nothing about the level of agreement of the actual observed measurements (Altman & Bland, 1983; Bland & Altman, 1986). Differences in scale (i.e., different means) between two methods clearly affect their agreement but do not change correlations. Regardless of the absolute magnitude of the error score variance, if it is small relative to true score variance then a high correlation will result and vice versa. When assessing agreement between two methods, we are really interested in a) the difference, on average, between measurements taken by the two techniques (i.e., bias), and b) the variation in the differences between measurements taken by the two techniques (i.e., limits of agreement; Altman & Bland, 1983; Bland & Altman, 1986, 1999). The amount of acceptable disagreement depends on the use of the measurement but should generally be defined before measurement (Bland & Altman, 1999).

The objectives of this study were to a) examine the reliability (i.e., relative and absolute) of repeated measurements, by flexible electrogoniometry (EG), of knee and ankle joint angle during walking and b) examine the concurrent validity (i.e., association and agreement) of EG and a previously established measure, 3D motion analysis (3D). It was hypothesized that

(1) EG measures taken from the same participant a) by two different testers successively at the same visit and b) by the same tester at two separate visits within 48 hours would be highly reliable (i.e., inter- and intra-tester reliability, respectively),

(2) the precision (i.e., measurement error) of EG would be similar to that reported by the majority of studies measuring gait kinematics using 3D,

(3) EG measures would be associated with 3D measures, and

(4) EG measures would agree with 3D measures.

2.2. Methods

The current study was approved by the University of Saskatchewan Behavioural Research Ethics Board (Appendix A).

2.2.1. Participants

Healthy adults were recruited from the undergraduate and graduate student population in the College of Kinesiology at the University of Saskatchewan. Participants were included if they were free from any disease or injury that would affect their normal walking gait.

2.2.2. Measures

2.2.2.1. Demographics

All participants' height and weight were measured on the same stadiometer and scale.

2.2.2.2. Flexible electrogoniometry (EG)

EG was used to measure joint angles at both knees (SG110, Biometrics Ltd., UK) and ankles (SG110/A, Biometrics Ltd., UK) during walking at a self-selected comfortable speed. A portable DataLOG Data Acquisition Unit (W4X8, Biometrics Ltd., UK) was connected via cable to each electrogoniometer and attached to a belt worn by the participant. Each electrogoniometer consists of two plastic endblocks connected by a wire inside a protective spring (Biometrics, 2007b). A series of strain gauges is mounted around this wire. As the relative angle between the endblocks changes, the corresponding change in strain on the connecting wire is measured and converted to a raw value between -4000 and 4000 units representing 180º rotation in each direction about the axis of rotation. The electrogoniometers are designed to attach to the body with the proximal endblock aligned to the body segment above the joint of interest and the distal endblock aligned to the body segment below. In this configuration, the measurements made by the electrogoniometers are an approximation of the angle of the body segments relative to one another (i.e., the joint angle). Each electrogoniometer contains two sets of strain gauges corresponding to the two axes perpendicular to the length of the endblocks allowing joint angle to be measured in two planes simultaneously, usually sagittal and frontal. Raw values are transmitted via wire from the electrogoniometer to the DataLOG unit where they are stored in a proprietary file on a removable memory card and simultaneously transmitted via Bluetooth to a nearby laptop where they can be monitored in real time and saved (Biometrics, 2007a). Measurements by flexible electrogoniometry in this study were sampled at 200 Hz and stored as raw values on a removable memory card in the DataLog unit.

2.2.2.3. 3D motion analysis (3D)

3D was used as the comparison criterion to which the validity of EG was tested. An infrared camera system (Vicon Nexus, Vicon Motion Systems, CO) was used to capture the coordinates of passive reflective markers attached to body segments while the participants walked. The system used eight infrared cameras around a 6-meter raised walkway and 54 reflective markers (i.e., 44 motion capture markers, 10 calibration markers) on the lower limbs of the participants (Figure 2.1). The X, Y, and Z coordinates of each marker relative to a constant global origin near the center of the walkway were sampled at 100 Hz for each calibration and walking trial and saved to a database.

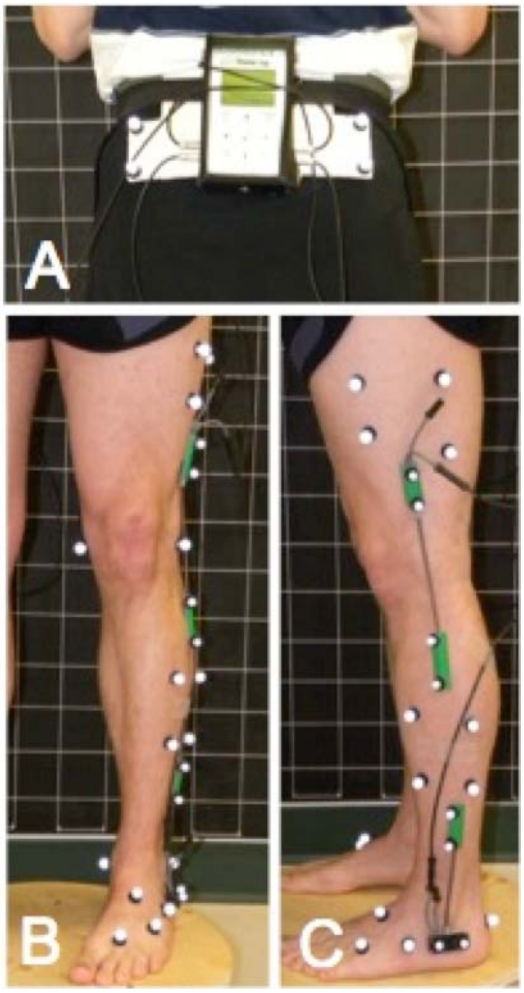

Figure 2.1. Three-dimensional motion analysis marker configuration. (A) Posterior view of pelvis markers. (B) Anterior view of leg markers including temporary configuration markers. (C) Lateral view of leg markers excluding temporary configuration markers.

2.2.3. Procedure

Each participant visited the lab on two occasions within 48 hours completing two sessions of walking trials at each visit as described below.

2.2.3.1. Visit 1

At the first visit, the participant was asked to provide consent to participate in the study (Appendix B) and demographic information was collected. Joint angles were measured in two separate walking sessions at each visit.

In the first session, one of two testers was randomly chosen to be the first tester to attach the electrogoniometers to the joints of the participant. For the next participant, the order of testers was reversed and then it was randomly chosen again for the following participant and so on. Each tester used the same standardized procedure to attach each electrogoniometer to each participant at each visit (Moriguchi et al., 2007; Piriyaprasarth et al., 2008; Rome & Cowieson, 1996; Rowe et al., 2001). Anatomical landmarks were located to ensure proper alignment of the electrogoniometer endblocks along the relevant body segments of the participant. Small round stickers were placed on each landmark so they could be easily removed and would not be visible to the next tester. The lateral malleolus, head of the fibula, and lateral condyle of the femur were landmarked while the participant was standing. The proximal and distal ends of the 5th metatarsal were landmarked while the participant was lying in the leg positioning blocks.

The leg positioning blocks ensured the participant's legs were in a uniform position while the tester attached and calibrated the electrogoniometers. The blocks consist of a 60 cm x 120 cm piece of wood that sits on the ground and a 60 cm x 30 cm piece of wood that is attached to the end of the first piece at a right angle. The end piece has a 20cm wide spacer block of wood attached to it. The participant was instructed to sit in the blocks with both legs extended, both heels firmly pressed into the angle of the two boards, both feet flat against the end board, and the insides of both feet pressed against the outside of the spacer block.

Electrogoniometers were then attached across the participant's knees and ankles using double-sided tape. On each ankle, the distal endblock was attached to the lateral side of the foot along an imaginary line extending out the posterior end of the 5^{th} metatarsal so that the coil wire was directly below the lateral maleolus. The proximal endblock was attached to the lateral side of the lower leg along an imaginary line between the head of the fibula and the tip of the lateral

malleolus. On each knee, the distal endblock was attached to the lateral side of the lower leg along an imaginary line between the head of the fibula and the tip of the lateral malleolus. The proximal endblock was attached to the lateral side of the upper leg along an imaginary line between the greater trochanter and the lateral condyle. During the placement of the proximal endblock the tester would palpate and landmark the greater trochanter with a thumb rather than marking with a sticker.

After attachment, the electrogoniometers were connected to the DataLog unit to calibrate them to 0˚ while still positioned in the leg positioning blocks as described earlier. Foot switches were then attached to the bottom of each heel and great toe of the participant making it possible to identify heel and toe contact events and delimit strides while the participant walks. The switches were held in place by tape and were then connected to the DataLog unit. The DataLog unit hung from a belt around the participant's waist so that it sat on top of the pelvis. All wires from the electrogoniometers and foot switches were then taped to the participant to ensure maximal comfort while walking. The 0º calibration, foot switch attachment, and wire management were always performed by the same individual.

Next, reflective markers were attached to the participant (Figure 2.1). The motion capture marker configuration consisted of four markers on a pelvis cluster attached to the back of a belt, four markers on each thigh, four markers on each shank, three markers on each forefoot, one marker on each heel, and two markers aligned along the axis of each electrogoniometer endblock. Temporary markers on the medial and lateral femoral epicondyles, medial and lateral malleoli, and second metatarsal head were used during calibration to mark the position of observable anatomical landmarks. Reflective 3D markers were always attached by the same individual.

Both the 3D and EG systems were then calibrated to a neutral position with the participant in quiet stance. Quiet stance was selected as the neutral position because of its balance of reproducibility and functionality (Moriguchi et al., 2007). Joint angle measurements were later calculated as the difference in joint angle from quiet stance. For 3D, this calibration also provided a static pose from which the position of the calibration markers relative to the walking trial markers could later be reconstructed. The participant was asked to stand in a comfortable stance with each foot on either side of a 20cm spacer block and both arms crossed with each hand on the opposite shoulder. The participant was asked to remain still in this

position for 5 seconds while a trial was recorded with both measurement systems (i.e., EG and 3D). The temporary 3D calibration markers were then removed before the participant performed a series of functional calibration trials for the hip and knee joints using only the 3D system.

For the hip functional calibration trials, the participant was asked to maximally flex and extend at the hip three times and then maximally abduct and adduct at the hip three times. For the knee, the participant was asked to maximally flex and extend the knee three times. The participant was then allowed 4-6 practice trials at a self-selected comfortable pace along the 6-meter raised walkway to ensure that all equipment was working properly and was attached comfortably.

Finally, the participant was asked to complete a series of ten walking trials at a self-selected comfortable pace along the 6-meter raised walkway. The following standard instructions were given to the participant, always by the same individual, to control for speed:

> *You're going to walk across the platform* [6-meter raised walkway] *at whatever pace you feel most comfortable and safe with. When you get to the other side of the platform, turn around and wait for our instruction to proceed again. We are going to do this 10 times. Ready? Start.*

After completing the walking trials, the electrogoniometers and landmarking stickers were removed by the first tester.

In the second session, the second tester replaced the electrogoniometers following the standardized procedure described above. The calibration markers were replaced as well and the procedure was repeated as described beginning with the calibration of both systems and ending with the participant completing another set of walking trials. After this second set of walking trials, the electrogoniometers, foot switches and markers were all removed and the participant was scheduled to return for a second visit within 48 hours.

2.2.3.2. Visit 2

At the second visit, the order of the testers placing the electrogoniometers was reversed (i.e., at the second visit, tester 2 attached the electrogoniometers on the participant first) and the participant was asked to complete another two sessions of walking trials as described above.

2.2.4. Data cleaning and reduction

Raw 3D output was first cleaned using Vicon Nexus for Windows (Vicon Motion Systems Ltd., Version 1.4.116) as follows. Leading and trailing frames of the 3D moving calibration trials (i.e., hip and knee) were trimmed to minimize the amount of time that the moving limb was stationary. 3D walking trials were trimmed to start at least one full gait cycle from gait initiation and include at least one full gait cycle. 3D moving calibration and trial data were cleaned to fill any gaps in marker visibility.

The remaining data reduction to stride variables was performed using MATLAB R2006b for Windows (The MathWorks, Inc., Version 7.3.0.267) as follows. Each body segment (i.e., thigh, shank, foot) was assigned an anatomical coordinate system based on observable (i.e., epicondyles and malleoli) and approximated anatomical landmarks (i.e., joint centers and axes; Cappozzo, Catanni, Della Croce, & Leardini, 1995). The static and functional calibration trials defined the positions of these anatomical landmarks relative to the 4-marker cluster used to track the respective body segment (Cappozzo et al., 1995). From the static pose, the positions of the knee and ankle joint centers were estimated as the point midway between the markers on the medial and lateral epicondyles and malleoli respectively. From the hip and knee functional calibration trials, the position of the hip joint center and knee flexion-extension axis were calculated (Ehrig, Taylor, Duda, & Heller, 2006; O'Brien, Bodenheimer, Brostow, & Hodgins, 2000). The position of the knee joint center was then relocated to the point on the knee flexion-extension axis closest to the knee joint center estimated from the static pose (Hagemeister et al., 2005). For each frame of each walking trial, the anatomical coordinate system for each body segment was reconstructed relative to the position of the respective 4-marker cluster. The relative 3D angles between these coordinate systems were calculated using a Cardan rotation sequence in which the flexion-extension angle was calculated first. Knee and ankle joint angles were defined by the flexion-extension angle between the thigh and shank coordinate system and the shank and foot coordinate system respectively. These angles were filtered using a second-order dual-pass low-pass Butterworth filter (8 Hz). Relative angles of the endblocks of the EGs were also determined from the coordinates of the markers on the endblocks and then filtered.

Raw EG values were downsampled to 100Hz by eliminating every second frame and filtered using a second-order dual-pass low-pass Butterworth filter (20 Hz) before being converted to joint angles. Each EG trial was then synchronized to the corresponding 3D trial by

matching the frame of peak ankle plantar-flexion measured by the angle of the 3D markers on the endblocks of the EG during the first swing phase in the 3D trial to the frame of peak ankle plantar-flexion during the same swing phase in the EG trial. Both ends of the EG trial were then cut to match the 3D trial.

Because of the potential clinical use of this measure in addition to its use in research, it is important that the outcome variables being assessed are comparable to those that would be used in a clinical setting (Bland & Altman, 1999; McGinley et al., 2009; Shrout & Fleiss, 1979). Thus, rather than analyze the entire waveform of the gait cycle, specific dependent variables were chosen because they a) had clinical relevance to MS (i.e., indicative of slow, short, stiff-legged steps), b) have been shown previously to be abnormal in people with MS using 3D, and c) provide examples of both range of joint angle motion (i.e., excursion) and discrete joint angle (i.e., at initial contact) measurements. From the ten trials for each participant at each visit, the last five usable strides were selected to provide a sample of the participant's walking[1]. Knee angle at initial contact, total knee excursion, ankle angle at initial contact, and total ankle excursion as measured by each device were calculated for each stride. Stride velocity as measured by 3D was also calculated for each stride because gait velocity is known to influence a wide range of gait characteristics (Bejek, Paróczai, Illyés, & Kiss, 2006; Kirtley, Whittle, & Jefferson, 1985; Stoquart, Detrembleur, & Lejeune, 2008). For each variable, the mean of the five strides was calculated.

2.2.5. Analytic plan

Mean and standard deviation were calculated to describe all demographic, velocity, and joint angle measurement variables. All variables from all participants at all time points (i.e., each session) were examined for missing data, and normality.

The first and second hypotheses, that a) EG measurements would show high inter- and intra-tester reliability and b) EG precision would be similar to those reported by the majority of studies measuring gait kinematics using 3D, were tested using the methods described by Eliasziw et al. (1994) on each dependent variable. This method uses all the EG measurements taken by

[1] This sample of 5 strides would be similar to what would be collected on a single trial across the 7-meter GAITRite walkway as will be described in Study 2 (i.e., examination of measures in a clinical sample).

both testers at both visits to concurrently calculate inter- and intra-tester reliability coefficients ($\hat{\rho}$), standard error of measurement (*SEM*), and minimum difference (*MD*) required to be fairly certain of real change. The reliability coefficients described by Eliasziw et al. (1994) are based on ICCs described by Shrout and Fleiss (1979).

EG measurements were entered into a repeated-measures analysis of variance (ANOVA) with tester and visit as factors. Tester was treated as a random effect so the results could be generalized to other testers. Estimates of variance components were calculated by equating expected mean squares to the observed mean squares from this ANOVA. Inter- and intra-tester reliability coefficients were calculated using the variance component estimates to form ratios of true score variance to observed score variance. To test the absolute reliability (i.e., precision) of EG, inter- and intra-tester *SEM* (i.e., the standard deviation of the error scores) were calculated as the square root of the variance components representing error score variance. The minimum difference (*MD*) required to signify real change was also calculated to allow clinicians and researchers to decide whether flexible electrogoniometry is appropriate for their specific clinical application or research question (Eliasziw et al., 1994).

The third and fourth hypotheses, that EG would be both associated with and in agreement with 3D (i.e., an established technique), were tested using Pearson correlation coefficients (*r)* to test association (Howell, 2007) and the methods described by Bland and Altman (1986; 1999; Altman & Bland, 1983) to test agreement for each dependent variable. Because repeated measures (i.e., four sessions) were obtained for each device on each participant, the mean of the four sessions was calculated for each device on each participant and used to compare the devices. The Pearson correlation coefficient (*r)* between EG and 3D was calculated to determine the level of association between measurements by the two devices. For each participant, the difference between the two devices was then calculated (i.e., EG - 3D) and checked for normality. The mean of the measurements by the two devices was calculated and used as the best estimate of the 'true' value of the variable. A Spearman rank correlation (r_s) was performed to formally test the relationship between absolute difference and mean of the two devices. The mean difference ($\bar{d}$) between the two devices was calculated to estimate bias of EG compared to 3D (i.e., if the mean difference is negative then EG tends to underestimate compared to 3D). The standard deviation of the differences was calculated, corrected to account for repeated measurement error, and used to construct 95% limits of agreement (*LOA*).

Differences in mean stride velocity between visits and between sessions within each visit were assessed using a repeated-measures analysis of variance (ANOVA) with Visit and Session as factors. Session was chosen as a factor rather than tester because stride velocity was assessed by 3D and instructions were always given by the same individual so stride velocity should not be influenced by which tester attached the electrogoniometers.

All statistical analyses were conducted using SPSS 13 for Mac OS X (SPSS Inc., Version 13.0.0). The level of significance for all statistical tests was set at 0.05.

2.3. Results

2.3.1. Participants and variable summary

Ten healthy adults (five males and five females) consented to participate in this study. Due to equipment malfunction, knee joint angle data (i.e., knee excursion and knee angle at initial contact) were missing from one testing session for one participant. Some of the variables were slightly skewed or kurtotic but the repeated-measures analyses used are sufficiently robust to smaller violations of normality (Howell, 2007). The mean and standard deviation of demographic variables from all participants are summarized in Table 2.1. The mean and standard deviation of joint angles measured with 3D and EG are summarized for each session in Table 2.2. At both the knee and ankle, 0º is when the participant is in quiet stance; positive values indicate knee flexion and ankle dorsi-flexion and negative values indicate knee extension and ankle plantar-flexion (Winter, 1991).

Table 2.1

Demographic variables of participants (N = 10).

Demographic variable	Mean	S.D.
Age (years)	25.61	3.02
Height (cm)	172.34	7.44
Weight (kg)	77.35	13.63

Table 2.2

Joint angles measured with 3D motion analysis (3D) and flexible electrogoniometry (EG) by each tester at each visit.

		Tester 1				Tester 2			
		Visit 1		Visit 2		Visit 1		Visit 2	
Joint angle		Mean	S.D.	Mean	S.D.	Mean	S.D.	Mean	S.D.
Knee excursion (degrees)	*3D*	66.30	5.27	68.31	5.17	66.42	6.14	68.29	5.73
	EG	63.29	5.22	65.08	7.13	62.29	5.52	62.44	5.72
Knee at initial contact (degrees)	*3D*	7.65	4.55	8.49	2.66	6.62	3.85	7.28	2.76
	EG	7.85	3.44	9.15	3.43	8.67	4.20	8.26	2.11
Ankle excursion (degrees)	*3D*	26.58	4.33	28.89	2.68	27.13	5.08	28.23	3.94
	EG	21.35	2.08	22.96	2.80	23.15	3.90	23.75	3.97
Ankle at initial contact (degrees)	*3D*	0.57	2.73	1.29	2.45	0.64	2.81	-0.06	2.18
	EG	2.31	2.77	1.94	2.98	1.22	2.22	1.01	3.09

2.3.2. Reliability of flexible electrogoniometry

The reliability of joint angles measured by flexible electrogoniometry during walking was assessed both relatively by reliability coefficients ($\hat{\rho}$) and absolutely by measurement error (*SEM*). Inter- and intra-tester $\hat{\rho}$ and *SEM* were calculated concurrently using variance components estimated from the observed mean squares of a repeated-measures ANOVA (Eliasziw et al., 1994) and are reported in Table 2.3. Generally poor inter- and intra-tester reliability was demonstrated for all variables ($\hat{\rho}_{inter}$ = 0.39 – 0.73, $\hat{\rho}_{intra}$ = 0.15 – 0.75). As would be expected, the findings indicate that intra-tester reliability is better than inter-tester reliability at the ankle; however, inter-tester reliability was better than intra-tester reliability at the knee. Inter- and intra-tester measurement errors were less than 3.5º for all variables (SEM_{inter} = 1.50 – 3.04, SEM_{intra} = 1.41 –3.19). The minimum difference (*MD*) required to signify real change was also calculated and is reported in Table 2.3 to allow clinicians and researchers to decide whether flexible electrogoniometry is appropriate for their specific clinical application or research question.

Table 2.3

Inter- and intra-tester reliability coefficients and standard error of measurement for joint angles measured by flexible electrogoniometry during walking.

	Inter-tester			Intra-tester		
Joint angle	$\hat{\rho}_{inter}$	SEM_{inter}	MD_{inter}	$\hat{\rho}_{intra}$	SEM_{intra}	MD_{intra}
Knee excursion	0.73	3.04	8.41	0.70	3.19	8.85
Knee at initial contact	0.50	2.38	6.60	0.15	3.11	8.63
Ankle excursion	0.39	2.63	7.29	0.51	2.35	6.51
Ankle at initial contact	0.72	1.50	4.15	0.75	1.41	3.92

2.3.3. Concurrent validity of flexible electrogoniometry and 3D motion analysis

The concurrent validity of EG and 3D was examined by assessing both their association (Pearson *r)* and their agreement (Altman & Bland, 1983; Bland and Altman 1986; 1999; Table 2.4). Bias of EG compared to 3D was assessed by the mean difference ($\bar{d}$) between the two devices. Accordingly, EG appears to underestimate both total knee and ankle excursion compared to 3D (knee excursion: $\bar{d}$ = -4.23; ankle excursion: $\bar{d}$ = -4.91). The limits of agreement

(*LOA*) indicate that, for all variables, we can expect EG measurements to be within 4º to 8º of 3D measurements 95% of the time. When measuring ankle excursion, the difference between the two devices was related to the mean of the two devices ($r_s = 0.72$, $p = 0.019$) so the limits of agreement will actually be slightly narrower than reported for smaller ankle excursions and slightly wider for larger ankle excursions.

Table 2.4

Mean, standard deviation, Pearson correlation coefficients and Bland and Altman limits of agreement for joint angles measured by flexible electrgoniometry and 3D motion analysis during walking.

	EG		3D		Association		Agreement	
Joint angle	Mean	SD	Mean	SD	*r*	*p*	$\bar{d}$	95% *LOA*
Knee excursion	63.10	5.49	67.33	5.37	0.86	0.001	-4.23	±8.36
Knee at initial contact	8.42	2.46	7.51	2.76	0.86	0.001	0.91	±6.74
Ankle excursion	22.80	2.51	27.71	3.60	0.75	0.012	-4.91	±7.54
Ankle at initial contact	1.62	2.50	0.61	2.31	0.86	0.001	1.01	±4.19

Excursion at the knee and ankle was calculated by subtracting the minimum angle (i.e., peak knee extension, peak ankle plantar-flexion) from the maximum angle (i.e., peak knee flexion, peak ankle dorsi-flexion). To further explore the source of the bias in the knee and ankle excursion measurements the following steps were taken:

(1) the bias of EG compared to 3D when measuring peak knee flexion and extension and peak ankle dorsi- and plantar-flexion and

(2) the bias in ankle excursion measured by EG compared to the excursion of the relative angle of the ankle electrogoniometer endblocks measured by 3D was assessed. (Recall that there were two reflective markers placed along the axis of each endblock allowing measurement of their relative angle by 3D.)

For all variables, the difference between the two devices was not related to the mean of the two devices. EG underestimated both peak flexion ($\bar{d}$ = -2.87) and extension ($\bar{d}$ = 1.35) at the knee while it only underestimated peak plantar-flexion ($\bar{d}$ = 5.23) at the ankle. The excursion of the ankle electrogoniometer endblocks was measured by 3D and compared to the ankle excursion measured by EG; there was essentially no bias ($\bar{d}$ = -0.25). The limits of agreement indicate that 95% of the time the two devices will be within 6.42º. Differences in how the two angles were measured probably contribute to this error. EG measured the angle in one plane (i.e., the sagittal plane) independent of rotation in other planes (Biometrics, 2007b). Because there are only two markers on each endblock, 3D was not able to define a sagittal plane so it could only measure the single relative angle between the long axes of the endblocks (as defined by the marker pairs on each endblock). Error would result if movement existed in any plane other than the sagittal plane such as inversion-eversion which occurs throughout the gait cycle (Inman, Ralston, & Todd, 1981).

2.3.4. Velocity

Differences in mean stride velocity between visits and between sessions within each visit were assessed using a repeated-measures analysis of variance (ANOVA). There was a significant main effect of both visit [$F_{(1,9)}$ = 8.14, p = 0.019] and session [$F_{(1,9)}$ = 7.67, p = 0.022]. Participants walked faster in the second session and in the second visit. This did not translate into significantly different EG joint angle measurements as the repeated-measures ANOVA used for

estimating variance components for the reliability coefficients did not indicate a significant main effect of visit for any of the joint angle variables.

2.4. Discussion

Reliable and valid clinically practical measures of kinematic gait parameters are needed so clinicians can easily but confidently assess the gait of patients with different movement conditions to examine a variety of clinical and basic research questions. The current study sought to examine the inter- and intra-tester reliability and precision of repeated EG measurements of knee and ankle joint angles during walking. The study also examined the concurrent validity of these measurements taken by EG and 3D, a previously established measure. The findings suggest that EG is a reliable and valid measure of knee and ankle joint angles during walking; however, given the preliminary nature of this study and some limitations, these conclusions need to be qualified.

EG showed excellent absolute reliability, but lower relative reliability, compared to that reported for 3D. In a review of 3D reliability by McGinley et al. (2009), reliability indices comparable to the reliability coefficients calculated in the current study were typically reported to exceed 0.80 for lower body joint angles measured in the sagittal plane at the knee and ankle during walking; absolute reliability (i.e., precision or measurement error) comparable to the *SEM*s calculated in the current study were typically reported to be lower than 4º for the same measurements. While none of the joint angle measurements by EG in the current study reached that level of relative reliability, all joint angle measurements had less than 4º of measurement error. The reliability coefficients may have been lower than expected for a number of reasons. In the current study, they may have been depressed by low between-participant variability relative to the amount of measurement error (Harvill, 1991; Weir, 2005). For this very reason, it is important to assess both relative and absolute reliability in this type of study (Harvill, 1991; McGinley et al., 2009; Weir, 2005).

As would be expected, the current findings indicate that, for EG, intra-tester reliability is better than inter-tester reliability at the ankle; however, inter-tester reliability is surprisingly better than intra-tester reliability at the knee. This finding may be explained by the presence of negative estimates of some variance components, which also affected inter-tester SEM. Using a

conservative approach, recalculating the affected variables using the absolute value of the variance components would not change the interpretation of the findings in this study.

Measurements by EG showed good association with measurements by 3D, especially considering the homogeneity of the sample since low between-participant variance is known to depress the correlation coefficient (Altman & Bland, 1983). However, it is not surprising that the two techniques are related considering they were both designed to measure the same quantity (Bland & Altman, 1986). Moreover, agreement between the measures was also good; the limits of agreement were similar to those reported for repeated measurements using 3D (Maynard, Bakheit, Oldham, & Freeman, 2003). However, EG did show some bias, underestimating both ankle and knee excursion by 4° to 5º on average due to underestimation of both peak extension and flexion at the knee and peak plantar-flexion at the ankle. There was no bias when the excursion of the ankle endblocks measured by 3D was compared with the ankle excursion measured by EG so the biases shown were probably not due to instrumental error but rather to some other factor affecting the actual position of the electrogoniometer endblocks such as soft tissue artifact (Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996), kinematic crosstalk (Rowe et al., 2001) or a combination of these factors.

Both EG and 3D measure the relative angles of body segments using devices attached to the external surface of the body that estimate the position of the underlying rigid bone (Cappozzo et al., 1995). During movement, change in passive and active soft tissue separating the measurement device from the underlying bone results in soft tissue artifact (Cappozzo et al., 1996; Peters, Galna, Sangeux, Morris, & Baker, 2010). A standardized procedure was used for placement of the endblocks resulting in very little measurement error between visits and between testers for EG measurements; however, differences in the placement of endblocks and reflective markers may be the cause of the bias reported. Due to their attachment locations, some EG endblocks may be especially prone to soft tissue artifact. The proximal endblock of the knee electrogoniometer is placed laterally, just superior to the knee, and may be affected by the contraction and relaxation of the vastus lateralis (Rowe et al., 2001). The distal endblock of the ankle electrogoniometer is placed inferior to the lateral malleolus and may be affected by compression (Gefen, Megido-Ravid, & Itzchak, 2001) and elastic rebound (Bennett & Ker, 1990) of the heel pad.

Kinematic crosstalk occurs when movement in one degree of freedom affects the measurement of movement in another degree of freedom (Rowe et al., 2001). Flexion-extension angles measured by EG are not sensitive to translation or internal-external rotation of the endblocks but are somewhat sensitive to abduction-adduction movement (Rowe et al., 2001). Up to 25º of ankle adduction (i.e., inversion; Moriguchi et al., 2007) occurring during peak ankle plantar-flexion, which was also observed as high as 25º in the current study, would still produce crosstalk errors of less than 2º (Rowe et al., 2001). Minimal abduction-adduction occurs at the knee but the attachment location of the proximal endblock of the knee electrogoniometer may again be problematic because the muscle bulk of the vastus lateralis creates a relative abduction angle between the proximal and distal endblocks. Although the magnitude of this angle is unknown and varies between individuals, when combined with up to 60º of knee flexion it may produce crosstalk errors close to the magnitude of the bias shown in this study (Rowe et al., 2001). Therefore, the bias between EG and 3D in this study is probably a combination of both soft tissue artifact and kinematic crosstalk.

Another concern was that the bias in the ankle excursion increased as the total excursion increased. However, this relationship disappeared when only examining peak dorsi- and plantar-flexion. Also, the bias of the EG did not appear related to the magnitude of the joint angle when considering different peaks at different joints. For example, peak plantar-flexion, which had a range of approximately -10º to -20º based on the mean of the two devices, showed the greatest amount of bias while peak ankle dorsi-flexion showed no bias at all and ranged from 7º to 12º. Peak knee flexion ranged from 57º to 72º and showed only slightly more bias than peak knee extension which ranged from -7º to 8º. This evidence does not support the notion that the amount of bias is related to the magnitude of the joint angle.

The present study found that knee and ankle joint angle measurements made by EG during walking contain very little measurement error when a standardized protocol is used to place electrogoniometer endblocks. Measurements by EG were also correlated with and in agreement with measurements taken concurrently by 3D. These findings support the use of EG as a reliable and valid measure of knee and ankle joint angles during walking and, along with its portability and ease-of-use, make EG an ideal candidate for assessing gait in a clinical setting. The present study used a standardized procedure for electrogoniometer endblock attachment and future studies should attempt to refine this procedure in order to explore and reduce the potential

effects of soft tissue artifact and kinematic crosstalk. Finally, the present study did not match the filter cut-off rates of EG and 3D data and it is recommended that this is done, if possible, in future studies to ensure that the signal and noise of both sets of measurements are attenuated in the same manner (Winter, 2005).

3. Study 2: Detecting kinematic gait abnormalities in multiple sclerosis using clinically practical measures

3.1. Introduction

The effects of MS on the CNS are often manifest as abnormalities in gait kinematics. People with MS tend to walk with reduced speed (Benedetti et al., 1999; Gehlsen et al., 1986; Givon et al., 2009; Gutierrez et al., 2005; Holden, Gill, & Magliozzi, 1986; Martin et al., 2006; Morris, Cantwell, Vowels, & Dodd, 2002; Orsnes, Sorensen, Larsen, & Ravnborg, 2000; Thoumie, Lamotte, Cantalloube, Faucher, & Amarenco, 2005), cadence (Benedetti et al., 1999; Givon et al., 2009; Holden et al., 1986; Thoumie et al., 2005), step length (Givon et al., 2009; Gutierrez et al., 2005; Holden et al., 1986) and stride length (Benedetti et al., 1999; Gehlsen et al., 1986; Gutierrez et al., 2005; Holden et al., 1986; Martin et al., 2006; Morris et al., 2002; Orsnes et al., 2000; Thoumie et al., 2005), and spend relatively less time in swing (Gutierrez et al., 2005) and more time in stance (Gutierrez et al., 2005; Orsnes et al., 2000) and double support (Benedetti et al., 1999; Gutierrez et al., 2005; Martin et al., 2006; Orsnes et al., 2000) compared to non-MS controls when walking at a self-selected comfortable pace. There is also some evidence that the joint angles of people with MS differ from controls at various points throughout the gait cyle. At initial contact, their hip and knee are more flexed (Benedetti et al., 1999) and their ankle is more plantar-flexed (Benedetti et al., 1999; Martin et al., 2006). At toe off, their hip and knee are still more flexed (Benedetti et al., 1999). During the swing phase, people with MS achieve a higher maximum hip flexion and a lower maximum ankle plantar-flexion (Benedetti et al., 1999). Over the entire gait cycle, people with MS tend to have an increased total excursion at the hip (Benedetti et al., 1999) and a decreased total excursion at the knee (Gehlsen et al., 1986) and ankle (Benedetti et al., 1999; Gehlsen et al., 1986). These abnormalities in people with MS result in a gait pattern that is characterized by slow, short, stiff-legged steps.

While these studies indicate the presence of kinematic gait abnormalities in people with MS they also have their limitations. Only four studies compared MS participants to a group of age- and gender-matched controls (Givon et al., 2009; Martin et al., 2006; Morris et al., 2002; Thoumie et al., 2005); the rest either matched for gender alone, or simply compared their results to normal gait characteristics found in the literature. As people age, it has been shown that their

gait pattern adapts to minimize challenges to balance (Winter, 1991). As with other autoimmune diseases, women are diagnosed with MS more often than men; however, men often have less favourable long-term outcomes (Deshpande, Kremenchutzky, & Rice, 2006). Thus, it is important to compare to an age- and gender-matched control group to be certain that differences are due to MS and not some other factor.

As well, only three studies reported measures of lower extremity joint angles (Benedetti et al., 1999; Gehlsen et al., 1986; Martin et al., 2006) and only one of these used age- and gender-matched controls (Martin et al., 2006). Measures of joint angle during walking are important because they provide additional description of the effects of MS on gait kinematics that are not noticed by the standard neurological evaluation or by other clinical tests of function (Benedetti et al., 1999; Martin et al., 2006). Finally, most of these studies used laboratory-based objective measures (i.e., video-motion analysis, instrumented treadmill) that are expensive and require expertise, time, and permanent space that may not be practical for all researchers or clinicians (Krebs et al., 1985). This is an important consideration if gait analysis of multiple sclerosis patients is to occur in a clinical setting to examine either basic research questions (e.g., about the control of gait via the central nervous system or the effects of multiple sclerosis on the central nervous system) or clinical questions (e.g., about the diagnosis, progression, and treatment of multiple sclerosis or movement disorders associated with it). According to the Multiple Sclerosis Clinical Outcomes Task Force, for a measure to be valuable clinically it must not only be reliable, valid, and sensitive to change in MS but also be practical; that is, it must be easy to administer, acceptable to both patients and health care professionals, and resource efficient (Rudick et al., 1996).

Although many technologies exist for assessing walking characteristics, two of note for their potential clinical portability are the GAITRite walkway system and flexible electrogoniometry (EG). The GAITRite walkway system is an example of a reliable (Bilney et al., 2003; Menz et al., 2004 ; van Uden & Besser, 2004), valid (Bilney et al., 2003; Cutlip et al., 2000; McDonough et al., 2001; Webster et al., 2005), and clinically practical objective measure of temporal and spatial gait characteristics. It is quick, portable, relatively inexpensive, and simple when compared to other objective assessments (van Uden & Besser, 2004) and has recently been used to measure spatial and temporal gait parameters of people with MS compared to an age- and gender-matched control group reporting similar findings to those reported

previously (Givon et al., 2009). Similarly, EG is a clinically practical objective measure of lower body joint angle. It is lightweight, portable, simple, and inexpensive (Moriguchi et al., 2007; Piriyaprasarth et al., 2008; Rowe et al., 2001) and has recently been found to be both reliable and valid[2] but has not yet been used to assess the gait of MS patients.

Therefore, the objective of this study was to evaluate the feasibility of using the GAITRite walkway and flexible electrogoniometry systems to detect kinematic gait abnormalities in people with MS when compared to an age- and gender-matched control group without MS. A secondary objective was to examine the association between these gait abnormalities and the MS patients' level of neurological impairment, which is indicative of their current stage in the progression of the disease. It was hypothesized that

(1) these clinically practical measures would detect differences between people with MS and people without MS in kinematic gait characteristics that have been previously shown to differ using other valid and reliable measures and

(2) these gait characteristics would be correlated with the MS patients' level of neurological impairment.

3.2. Methods

The current study was approved by the University of Saskatchewan Behavioural Research Ethics Board (Appendix C).

3.2.1. Participants

A group of participants with MS and a control group of participants without MS were recruited. Participants with MS were recruited from patients at the MS Clinic at the Saskatoon City Hospital who agreed to be contacted by the researchers with information about the study. MS participants were included if they were at least 18 years old and had been clinically diagnosed with MS (McDonald et al., 2001; Polman et al., 2005); they were excluded if they were not ambulatory with or without an assistive device or if they had been diagnosed with any unrelated neurological, musculoskeletal, or cardiovascular condition affecting lower limb function. Age- and gender-matched participants without MS were recruited by word-of-mouth through University staff, faculty, and students to serve as controls. Control participants were

[2] Reliability and validity of flexible electrogoniometry were examined and discussed in Study 1.

excluded if they were not ambulatory with or without an assistive device or if they had been diagnosed with any unrelated neurological, musculoskeletal, or cardiovascular condition affecting lower limb function.

3.2.2. Measures

3.2.2.1. Interview (Appendix D)

Each participant's height and weight was measured on the same stadiometer and scale. As well, the length of each participant's left and right legs was measured from the greater trochanter to the floor while standing with feet together. Participants were then interviewed to determine their date of birth, gender, dominant hand and additional information about their personal medical history including diagnoses of other neurological conditions, cardiovascular disease, musculoskeletal problems, etc.; surgical history; family medical history of MS and other neurological conditions; and MS-specific medical history if in the MS group.

3.2.2.2. Expanded Disability Status Scale (EDSS; Appendix E)

The EDSS is a measure of impairment due to neurologic dysfunction caused by MS; it is based upon the standard neurologic examination (Kurtzke, 1983). In the standard neurologic examination, a patient is assigned a numerical grade for each of the following Functional Systems: Pyramidal, Cerebellar, Brain Stem, Sensory, Bowel and Bladder, Visual, Cerebral, and Other. Based on the combination of grades in these Functional Systems, the patient is assigned an EDSS score between 0 (normal) and 10 (death due to MS). Scores from 1 to 10 increase by 0.5 increments. From EDSS 0 to 3.0, the score is based entirely on Functional System grades. From EDSS 3.5 to 5.5, these grades are considered in combination with the patient's ability to perform daily activities (e.g., work) and their ambulatory status (i.e., distance travelled with or without assistive devices). EDSS scores from 6.0 to 10 are based almost entirely on ambulatory status (i.e., wheelchair use) and self-care function (e.g., eating).

3.2.2.3. Flexible electrogoniometry

Flexible electrogoniometry, along with its reliability and validity, is discussed in detail in Study 1.

3.2.2.4. GAITRite walkway system

The GAITRite instrumented walkway system (GAITRite Platinum v3.9, CIR Systems, Inc., Havertown, PA) includes a thin mat embedded with pressure sensors allowing participants to walk freely while temporal and spatial data is quickly and easily collected and transmitted to proprietary software on a connected computer (CIR Systems, Inc., 2007). The pressure sensors are placed on 1.27 cm centers arranged into a 61 cm x 61 cm pad so each pad is a 48 x 48 grid of sensors totaling 2304 sensors. These pads are then placed in series and embedded into an open cell foam rubber mat with water and chemical resistant vinyl top cover; the mat is very portable and available in a variety of lengths ranging from an active area of 366 cm to 1098 cm. The mat in the current study had an active length of 610 cm and was comprised of 23040 sensors (i.e., 48 x 480). The actual mat encasing the sensors is slightly larger than the active area; the mat in the current study was 700 cm long by 91 cm wide. Temporal and spatial gait measurements from the GAITRite walkway were sampled at 120 Hz.

3.2.3. Procedure

Upon arrival at the laboratory, each participant was asked to provide consent to participate in the study and release access to his or her most recent MS-related neurological assessment (i.e., EDSS; Appendix F). The participant was then interviewed briefly to obtain relevant demographic information and additional information about his or her disease state.

Next, the participant performed a series of walking trials across the GAITRite electronic walkway with flexible electrogoniometers attached to both knees and ankles. Using the same standardized protocol described previously[3], electrogoniometers were attached to the participant's knees and ankles, the EG system was calibrated to 0º, foot switches were attached to the bottom of each heel and great toe of the participant, all electrogoniometers and foot switches were attached to the DataLog unit, and the EG system was calibrated to quiet stance (QS). The participant was then allowed 2-3 minutes of walking about the room to ensure all equipment was working properly and attached comfortably.

Next, the participant was asked to complete a series of three walking trials at a self-selected comfortable pace along an 11-meter walkway. The walkway included start and finish

[3] The standardized protocol used to attach the electrogoniometers is described in Study 1 under section *2.3. Procedure.*

lines marked 2 m beyond either end of the GAITRite walkway. The following standard instructions were given to the participant to control for speed:

> *You're going to walk across the mat at whatever pace you feel most comfortable and safe with. Do not slow down until after you pass the tape at the other side. When you pass the tape at the other side, turn around and wait for our instruction to proceed again. We are going to do this 3 times. Ready? Start.*

During each walking trial, the first footfall to land on the GAITRite mat was noted (e.g., L2 indicates the second footfall of the left foot). After completing the walking trials, the electrogoniometers were removed.

3.2.4 Data cleaning and reduction

Using the GAITRite software, each walking trial of each participant was cleaned to verify the accuracy of each footfall. Leading or trailing footfalls that were only partially within the active measurement area were deleted and data was calculated based on only the verified footfalls. The first footfall in the active measurement area was verified to ensure it matched the first footfall on the mat. If these footfalls did not match, the first verified footfall was adjusted accordingly. For example, if the tester's notes stated L2 (i.e., second footfall with left foot) was the first footfall on the mat, the first verified footfall might be on the right foot because the preceding left footfall (L2) was only partially in the active area so it was deleted. The first verified footfall would therefore be adjusted to R2 or R3 based on their starting foot which could be derived from the EG data. For each walk, a wide range of variables was exported from the GAITRite software to a spreadsheet. Velocity, cadence, stride length, left and right step length, left and right stance (% of gait cycle), and left and right double support (% of gait cycle) from each walking trial were the spatial and temporal variables used in the current study.

Raw EG values were filtered using a second-order dual-pass low-pass Butterworth filter (20 Hz) before being converted to joint angles. Each EG trial was synchronized to the corresponding GAITRite walking trial by cutting the trial just before the initial contact corresponding to the first verified footfall on the GAITRite and just after the initial contact corresponding to the last verified footfall of the GAITRite. Knee angle at initial contact, total knee excursion, ankle angle at initial contact, and total ankle excursion were calculated for each stride of each trial. Data reduction from raw output to stride variables was performed using

MATLAB R2006b for Windows (The MathWorks, Inc., Version 7.3.0.267). For each joint angle variable, the mean of the strides for each trial was calculated in Microsoft Excel 2008 for Mac (Microsoft Corporation, Version 12.2.3).

For all gait variables, the mean over the last two walking trials was calculated in Microsoft Excel 2008 for Mac (Microsoft Corporation, Version 12.2.3).

3.2.5 Analytic plan

Mean and standard deviation were calculated to describe all demographic and kinematic variables for both the MS and control group. All variables were examined for normality. For each independent *t* test described below, Levene's test for equality of variance tested the variable for homogeneity of variance between the two groups. Independent *t* tests were used to test for differences between the two groups for each demographic variable.

To test the first hypothesis, that GAITRite and EG will detect differences between people with MS and non-MS controls in gait characteristics that have been previously shown to differ using other valid and reliable measures, independent *t* tests compared each kinematic gait parameter between the groups. Although there are multiple, correlated dependent variables obtained from the measures described, conducting an omnibus multivariate analysis would have been premature. This study has relatively low power for a multivariate test given the small sample size and multiple dependent variables. As well the study is exploratory in nature. Relying on omnibus test results alone when there is a null effect could result in making a Type II error and overlooking valuable information about the MS and non-MS comparisons of the various measures of gait. Given that conducting multiple *t* tests can increase the probability of Type I error (i.e., detecting effects due to chance alone), caution is used when interpreting the results of these tests (Howell, 2007).

To test the second hypothesis, that kinematic gait characteristics would be correlated with level of neurological impairment, Spearman correlation coefficients for ranked data (r_s) were calculated between EDSS score and each kinematic gait variable.

All statistical analyses were conducted using SPSS 13 for Mac OS X (SPSS Inc., Version 13.0.0). The level of significance for all statistical tests was set at 0.05.

3.3. Results

3.3.1 Demographics

Six female MS patients and six age- and gender-matched controls consented to participate in this study. The mean and standard deviation of all demographic variables is summarized in Table 3.1. There was no statistical difference between the two groups for any of the demographic variables. The EDSS and Functional System scores of each participant with MS are shown in Table 3.2.

Table 3.1

Demographic characteristic of MS patients and controls.

		MS (N = 6)		Control (N = 6)	
Characteristic		Mean	S.D.	Mean	S.D.
Age (years)		46.41	9.85	46.51	10.23
Height (cm)		167.20	8.39	166.35	4.17
Weight (kg)		78.83	24.03	59.83	11.46
Leg length (cm)	*L*	89.08	4.64	86.08	2.76
	R	89.67	4.76	86.42	3.35
EDSS		2.58	1.07	-	-

Table 3.2

MS participant EDSS and Functional System scores.

		Functional System							
Participant	EDSS	P	Cll	BS	S	BB	V	Cb	O
001	4.0	3	3	2	0	2	0	3	0
002	2.0	1	2	0	1	1	0	0	0
003	1.5	1	1	0	1	1	0	0	0
004	1.5	1	1	0	1	1	0	0	0
005	3.0	1	2	0	2	2	0	0	0
006	3.5	3	3	2	2	2	0	0	0

P = Pyramidal, Cll = Cerebellar, BS = Brain Stem, S = Sensory, BB = Bowel and Bladder, V = Visual, Cb = Cerebral, O = Other.

3.3.2 Gait characteristics

Gait characteristics of the two groups are presented in Table 3.3 with significant differences indicated. The MS group differed from the control group on a number of spatial and temporal gait characteristics. They had a significantly slower mean velocity with corresponding significantly lower cadence and showed a trend toward shorter stride and step length. The MS group spent a significantly greater percentage of the gait cycle in the stance phase and in double support. The only significantly different joint angle characteristic was ankle excursion on the left leg, which was significantly lower in the MS group.

Table 3.3

Gait characteristic of MS patients and controls.

		MS (N = 6)		Control (N = 6)		
Gait characteristic		Mean	S.D.	Mean	S.D.	p
Velocity (cm/sec)		99.73*	25.05	132.10	23.16	0.043
Cadence (steps/min)		108.03*	11.53	121.83	7.95	0.036
Stride length (cm)		110.08	20.15	129.40	16.24	0.097
Step length (cm)	*L*	55.74	9.33	64.37	7.85	0.113
	R	53.99	11.04	65.01	8.33	0.080
Stance (% gait cycle)	*L*	62.63*	1.79	59.03	2.05	0.009
	R	62.58*	1.81	59.48	1.54	0.010
Double support (% gait cycle)	*L*	25.22*	2.88	18.69	3.25	0.004
	R	25.30*	3.05	18.66	3.30	0.005
Knee excursion (degrees)	*L*	51.28	6.63	55.27	3.51	0.222
	R	49.80	5.33	52.66	2.57	0.265
Knee at initial contact (degrees)	*L*	4.64	10.15	5.02	4.17	0.934
	R	10.13	6.60	5.89	2.82	0.178
Ankle excursion (degrees)	*L*	15.84*	4.08	19.78	1.18	0.046
	R	18.69	4.22	19.35	2.29	0.740
Ankle at initial contact (degrees)	*L*	-0.18	3.81	-1.77	2.81	0.428
	R	-1.31	4.60	-3.28	1.76	0.350

* significantly different than contols ($p < 0.05$)

3.3.3 Association between gait characteristics and neurological impairment

Correlations between EDSS score and gait characteristics are presented in Table 3.4. None of the correlations were significant although there was a trend towards a positive correlation between the portion of the gait cycle spent in stance and EDSS score.

Table 3.4

Spearman correlations between EDSS and gait variables.

Gait variable		r_s	p
Velocity (cm/sec)		-0.32	0.538
Cadence (steps/min)		-0.35	0.499
Stride length (cm)		-0.35	0.499
Step length (cm)	*L*	-0.38	0.461
	R	-0.35	0.499
Stance (% gait cycle)	*L*	0.73	0.103
	R	0.23	0.658
Double support (% gait cycle)	*L*	0.46	0.354
	R	0.29	0.577
Knee excursion (degrees)	*L*	0.06	0.913
	R	0.12	0.827
Knee at initial contact (degrees)	*L*	0.20	0.700
	R	0.23	0.658
Ankle excursion (degrees)	*L*	0.32	0.538
	R	0.17	0.742
Ankle at initial contact (degrees)	*L*	0.15	0.784
	R	-0.38	0.461

3.4. Discussion

Objective measurement of kinematic gait characteristics in people with MS may be useful in the examination of both clinical and basic research questions. Usually, it is most practical for these assessments to occur in a clinical setting requiring instruments that are not only reliable and valid, but also clinically practical. Using clinically practical measures, this study confirmed a

number of kinematic gait abnormalities previously detected in people with MS using other valid and reliable objective measures.

Using GAITRite, the current study confirmed that MS patients walked slower than age- and gender-matched controls with a corresponding reduction in cadence and a trend toward reduction in step length and stride length. A simple explanation may be that people with MS are forced to reduce their walking speed because they have a higher metabolic cost of walking; that is, they expend more energy over a given distance than people without MS (Olgiati, Burgunder, & Mumenthaler, 1988). If people tend to pick a comfortable walking speed to optimize energy expenditure relative to distance (Ralston, 1958; Saunders, Inman, & Eberhart, 1953), then people with MS may select a slower speed to maintain optimal energy expenditure. To reduce speed, they would reduce both cadence and step/stride length (Andriacchi, Ogle, & Galante, 1977).

Using EG, the current study also confirmed that people with MS had a lower total ankle excursion while walking. Along with gait kinematics, Benedetti et al. (1999) and Martin et al. (2006) also examined muscle activity during gait and found that people with MS had increased co-activation of antagonistic muscles at the ankle corresponding to this reduction in excursion. In MS, co-activation may be a result of spasticity (Sinkjaer, Anderson, & Nielsen, 1996), which has been associated with the higher metabolic cost of walking in people with MS (Olgiati et al., 1988). This finding supports the possibility that people with MS walk slower than people without MS to maintain optimal energy expenditure. Some of the other joint angle variables (i.e., left knee excursion) appeared to differ between the groups; however, increased variability in the group of participants with MS compared to those without MS (Table 3.4) may have prevented the detection of significant effects. This increase in variability has previously been linked to fatigue (Albrecht et al., 2001; Crenshaw, Royer, Richards, & Hudson, 2006) but may also be linked to pain or motivation. The increase in variability in people with MS would be interesting to examine given a larger sample with greater statistical power than offered by the current study.

Finally, the current study confirmed that people with MS spent a larger portion of the gait cycle in the support phases of gait: stance and double support. In addition to kinematic parameters, Orsnes et al. (2000) measured ground reaction forces to calculate unsteadiness of gait and found that people with MS had higher gait unsteadiness than people without MS. Due to this unsteadiness, people with MS may choose a more conservative gait pattern featuring

increased time in the support phases of gait and a necessary reduction in speed (Benedetti et al., 1999; Gutierrez et al., 2005; Martin et al., 2006; Orsnes et al., 2000).

The current study found no statistically significant association between any of the kinematic gait variables and the MS patients' level of neurological impairment. This finding may be explained by both low power and the nature of EDSS, the measure of neurological impairment used in this study. EDSS relies more heavily on ambulatory status as patients move up the scale. Until level 3.0, EDSS is based entirely on Functional System grades, which may or may not be associated with gait impairment. From level 3.5 to level 6.5, ambulatory status with or without assistive devices is a major basis for distinction between levels. From level 7.0 upward, patients are essentially restricted to a wheelchair or bed. The EDSS score of MS participants in this study ranged from 1.0 to 4.0, below the range where ambulatory status plays a major role in determination of neurological impairment.

This study was limited by low sample size, and thus low statistical power, which may have hindered the detection of significant joint angle differences between the two groups. Seven to 15 participants were needed to achieve an 80% power of detecting joint angle differences with an expected effect size of 1 to 1.5 (Benedetti et al., 1999). In addition, low power may have limited the detection of significant correlations between EDSS and kinematic gait variables. At least 23 participants were required to have the same power of detecting medium ($r = \pm 0.5$) correlations between EDSS and the kinematic gait variables (Gehlsen et al., 1986; Givon et al., 2009).

Interpretation of the results of this study may be confounded by the influence of walking speed on other gait parameters. Specifically, each of the gait parameter differences between groups is influenced by speed of walking in healthy adults (Bejek et al., 2006; Kirtley et al., 1985; Stoquart et al., 2008). Therefore, the between-group differences could be due to a) disease characteristics among individuals with MS, b) slower walking speed among individuals with MS, or c) a combination of a) and b). Accordingly, while a number of possible reasons (i.e., increased metabolic cost due to spasticity, gait unsteadiness) for the kinematic gait abnormalities detected in people with MS (relative to healthy matched controls) have been suggested based on the results of previous studies, metabolic cost, muscle co-activation, spasticity, and gait unsteadiness were not measured in this study so their effects on gait kinematics in people with MS remain speculative.

As an exploratory investigation, the current study raised as many questions as it answered indicating a number of suggestions. Future research should include clinically practical measures of gait kinematics alongside clinically practical measures of metabolic cost, muscle co-activation, spasticity, and gait unsteadiness. Such a battery of measures would offer the opportunity to explore the relationships between these variables and further model the effects of MS on gait. In exploration of the increase in gait variability in people with MS, measures of fatigue (Albrecht et al., 2001; Crenshaw et al., 2006), pain and motivation may also be added. As well, researchers should be careful when attempting to examine the relationship between gait abnormalities and disease progression as measured by the most common scale, EDSS. If incorporating MS patients with an EDSS score of 3.5 to 6.5 in future studies, there may be a better association between these gait abnormalities and neurological impairment. Alternatively, it may be interesting to explore the relationship between kinematic gait abnormalities and other clinical assessments that incorporate a quantitative measure of gait. The Multiple Sclerosis Functional Composite (MSFC) includes a timed 25-foot walk (Fischer, Rudick, Cutter, & Reingold, 1999) and is predictive of both clinical and magnetic resonance imaging (MRI) status of patients with MS (Rudick et al., 2001). Finally, since GAITRite and EG are capable of detecting some kinematic gait abnormalities even in a small sample of modestly disabled MS patients, their use in clinical research with larger numbers of participants may prove fruitful as exemplified by a recent study using GAITRite (Givon et al., 2009).

The current study used GAITRite and EG to detect kinematic gait abnormalities in people with MS supporting the use of these clinically practical measures in the assessment of gait kinematics. People with MS walked slower by reducing both cadence and stride length, spent a greater portion of the gait cycle in stance and double support, and had a lower total joint excursion at the ankle than people without MS. Future research should explore the relationship between these gait abnormalities and features of MS such as neurological impairment, spasticity, and gait unsteadiness.

4. General discussion

While a handful of studies have described the kinematic gait pattern of people with multiple sclerosis (Benedetti et al., 1999; Gehlsen et al., 1986; Givon et al., 2009; Gutierrez et al., 2005; Holden et al., 1986; Martin et al., 2006; Morris et al., 2002; Orsnes et al., 2000; Thoumie et al., 2005), these studies have had their limitations including lack of age- and gender-matched controls, lack of joint angle measures and the use of laboratory gait analysis which may not be practical for clinicians or researchers who wish to perform research in a clinical setting. Clinically practical measures of gait kinematics, such as GAITRite and EG, may provide clinicians and researchers with a tool to assess gait in clinical populations in order to examine basic research and clinical questions. The objective of the studies in this research project was to explore whether clinically practical measures could be used to accurately detect kinematic gait abnormalities in people with multiple sclerosis. The findings of these studies indicate that EG is a reliable and valid measure of knee and ankle joint angles during walking that can be used along with GAITRite to detect kinematic gait abnormalities in people with MS.

Specifically, the first study found that measurements of knee and ankle joint angle taken by EG during walking contained little measurement error and were in agreement with measurements taken concurrently by 3D when a standardized protocol was used to place the electrogoniometer endblocks. Using both GAITRite and EG, the second study found that people with MS walked slower as a result of a lower cadence and possibly shorter steps, spent a greater portion of the gait cycle in the support phases of gait, and had reduced joint excursion at the ankle compared to age- and gender-matched controls. These findings support the use of GAITRite and EG in the examination of clinical research questions about gait in people with MS. Numerous research directions are immediately apparent based on the findings of the present studies.

In a clinical setting, GAITRITE and EG should be combined with clinically practical measures of gait kinetics, muscle activity, spasticity, and energy expenditure to examine the precise mechanisms behind the kinematic abnormalities that have been shown to exist in people with MS. A reduction in walking speed is a common finding of all previous studies of gait kinematics in people with MS (Benedetti et al., 1999; Gehlsen et al., 1986; Givon et al., 2009; Gutierrez et al., 2005; Holden et al., 1986; Martin et al., 2006; Morris et al., 2002; Orsnes et al., 2000; Thoumie et al., 2005). Walking speed is known to affect other kinematic gait

characteristics (Bejek et al., 2006; Kirtley et al., 1985; Stoquart et al., 2008) so it would be beneficial to determine which, if any, gait abnormalities are causes of the reduction in gait speed and which are effects of it. Two possible contributors to walking speed reduction in MS patients are an increase in the metabolic cost of walking due to spasticity (Olgiati et al., 1988; Ralston, 1958; Saunders et al., 1953) and an increase in gait unsteadiness (Benedetti et al., 1999; Orsnes et al., 2000). GAITRite and EG should be used to examine the relationship between gait kinematics and measures of these contributors.

It is possible that the specific mechanisms acting to produce gait abnormalities in people with MS, as with other signs and symptoms, may vary depending on the location of lesions in the CNS (Pryse-Phillips & Sloka, 2006). The most recent studies of gait kinematics in people with MS have attempted to characterize abnormalities in different subgroups distinguished by the presence of signs in the different functional systems of the EDSS (Givon et al., 2009; Martin et al., 2006; Thoumie et al., 2005). Thoumie et al. (2005) compared subgroups that had pyramidal tract signs either alone or in combination with sensory signs, cerebellar signs or both. They found that patients with sensory signs walked with a higher cadence than the other groups. Using GAITRite, Givon et al. (2009) determined a different pattern of gait impairment between people with pyramidal signs and people with cerebellar signs. Neither of these studies included a measure of joint angles. Martin et al. (2005) found that MS patients without pyramidal signs walked fast, took longer strides, and spent a smaller portion of the gait cycle in double support compared to patients with pyramidal signs. Martin et al. (2005) also measured knee and ankle joint angles using 2D motion analysis but found no differences between these two groups. A clinically practical measure of joint angle during walking, such as EG, should be included in future studies.

When assessing the gait of people with MS, there appears to be added value in the use of GAITRite alongside observational analysis by a clinician. As mentioned, previous research has shown the validity and reliability of GAITRite and the current study has shown its value in detecting kinematic gait abnormalities in people with MS. While observational analysis provides a rich description of the movement, objective measures of gait kinematics can detect abnormalities that might not otherwise be detected (Benedetti et al., 1999; Martin et al., 2006). Benedetti et al. (1999) argue that subclinical gait abnormalities should be seen as paraclinical evidence of central nervous system lesions to assist in the diagnosis of MS (Poser et al., 1983).

More realistically, the early identification of subclinical gait abnormalities by GAITRite should be used to identify movement disorders related to MS and to intervene with specific treatment (Martin et al., 2006). Use of GAITRite to dictate treatment need not be reserved for the onset of gait abnormality as its sensitivity also makes it ideal for tracking the progression of gait impairments associated with MS and assessing the efficacy of treatments (Martin et al., 2006).

The jury is still out, however, on the use of EG in clinical practice. The first study indicated that a change of between 4° and 8°, depending on the variable, was required to indicate a clinically significant change. In the second study, the only significant between-groups difference identified was mean left ankle excursion; however, the difference was less than 4°. While the EG may be sensitive enough to detect statistically significant differences between groups in a research study, it may not be sensitive enough to detect change over time in a patient with MS as the disease progresses. There are additional practical challenges to the use of EG in a clinical setting. As seen in study 1, care must be taken to follow a standardized procedure when attaching the electrogoniometers in order to minimize soft tissue artifact (Moriguchi et al., 2007; Piriyaprasarth et al., 2008; Rome & Cowieson, 1996; Rowe et al., 2001). This procedure can be time consuming and potentially fatiguing for more disabled participants. Future studies should explore improvements to the procedure of attaching electrogoniometers. For clinical research participants and for patients, improvements would reduce participant burden. For accuracy of EG assessment, improvements may reduce soft tissue artifact to improve the sensitivity of the measure.

References

Agraharasamakulam, S. N., Bronner, S., Ojofeitimi, S. (2005, August). Comparison of two ankle electrogoniometers and motion analysis. Paper presented at the meeting of the International Society of Biomechanics Annual Conference, Cleveland, OH.

Albrecht, H., Wötzel, C., Erasmus, L. P., Kleinpeter, M., König, N., & Pöllman, W. (2001). Day-to-day variability of maximum walking distance in MS patients can mislead to relevant changes in the Expanded Disability Status Scale (EDSS): Average walking speed is a more constant parameter. *Multiple Sclerosis, 7,* 105-109.

Altman, D. G. & Bland, J. M. (1983). Measurement in medicine: The analysis of method comparison studies. *The Statistician, 32,* 307-317.

Andriacchi, T. P., Ogle, J. A., & Galante, J. O. (1977) Walking speed as a basis for normal and abnormal gait measurements. *Journal of Biomechanics, 10,* 261-268.

Ashley-Ross, M. A. & Gillis, G. B. (2002). A brief history of vertebrate functional morphology. *Integrative and Comparative Biology, 42,* 183-189.

Bejek, Z., Paróczai, R., Illyés, Á., & Kiss, R. M. (2006). The influence of walking speed on gait parameters in healthy people and in patients with osteoarthritis. *Knee Surgery, Sports Traumatology, Arthroscopy, 14,* 612-622.

Benedetti, M. G., Piperno, R., Simoncini, L., Bonato, P., Tonini, A., & Giannini, S. (1999). Gait abnormalities in minimally impaired multiple sclerosis patients. *Multiple Sclerosis, 5,* 363-368.

Bennett, M. B. & Ker, R. F. (1990). The mechanical properties of the human subcalcaneal fat pad in compression. *Journal of Anatomy, 171,* 131-138.

Benoit, D. L., Ramsey, D. K., Lamontagne, M., Xu, L., Wretenberg, P., & Renström, P. (2006). Effect of skin movement artifact on knee kinematics during gait and cutting motions measured in vivo. *Gait & Posture, 24,* 152-164.

Bilney, B., Morris, M., & Webster, K. (2003). Concurrent related validity of the GAITRite® walkway system for quantification of the spatial and temporal parameters of gait. *Gait & Posture, 17,* 68-74.

Biometrics. (2007a). DataLog Bluetooth & MMC: Operating manual. Gwent, UK: Author.

Biometrics. (2007b). Goniometer and torsiometer operating manual. Gwent, UK: Author.

Bland, J. M. & Altman, D. G. (1986). Statistical methods for assessing agreement between two methods of clinical measurement. *The Lancet, i,* 307-310.

Bland, J. M. & Altman, D. G. (1999). Measuring agreement in method comparison studies. *Statistical Methods in Medical Research, 8,* 135-160.

Cameron M. H., Horak, F. B., Herndon, R. R., & Bourdette, D. (2008). Imbalance in multiple sclerosis A result of slowed spinal somatosensory conduction. *Somatosensory and Motor Research, 25,* 113-122.

Cappozzo, A., Catani, F., Della Croce, U., & Leardini, A. (1995). Position and orientation in space of bones during movement: anatomical frame definition and determination. *Clinical Biomechanics, 10,* 171-178.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., & Della Croce, U. (1996). Position and orientation in space of bones during movement: experimental artefacts. *Clinical Biomechanics, 11,* 90-100.

Chien, S. L., Lin, S. Z., Liang, C. C., Soong, Y. S., Lin, S. H., Hsin, Y. L., et al. (2006). The efficacy of quantitative gait analysis by the GAITRite system in evaluation of parkinsonian bradykinesia. *Parkinsonism and Related Disorders, 12,* 438-442.

CIR Systems, Inc. (2007). GAITRite manual version 3.9. Havertown, PA: Author.

Clarac, F. (2008). Some historical reflections on the neural control of locomotion. *Brain Research Reviews, 57,* 13-21.

Corradini, M. L., Fioretti, S., Leo, T., & Piperno, R. (1997). Early recognition of postural disorders in multiple sclerosis through movement analysis: A modeling study. *IEEE Transaction on Biomedical Engineering, 44,* 1029-1038.

Coutts, F. (1999). Gait analysis in the therapeutic environment. *Manual Therapy, 4,* 2-10.

Crenshaw, S. J., Royer, T. D., Richards, J. G., & Hudson, D. J. (2006). Gait variability in people with multiple sclerosis. *Multiple Sclerosis, 12,* 613-619.

Cutlip, R. G., Mancinelli, C., Huber, F., & DiPasquale, J. (2000). Evaluation of an instrumented walkway for measurement of the kinematic parameters of gait. *Gait and Posture, 12,* 134-138.

Deshpande, R., Kremenchutzky, M., & Rice, G. P. A. (2006). The natural history of multiple sclerosis. *Advances in Neurology, 98,* 1-16.

Ehrig, R. M., Taylor, W. R., Duda, G. N., Heller, M. O. (2007). A survey of formal methods for determining functional joint axes. *Journal of Biomechanics, 40,* 2150-2157.

Eliasziw, M., Young, S. L., Woodbury, M. G., & Fryday-Field K. (1994). Statistical methodology for the concurrent assessment of interrater and intrarater reliability: Using goniometric measurements as an example. *Physical Therapy, 74,* 777-788.

Ferdirchuk, B., Nielsen, J., Petersen, N., & Hultborn, H. (1998). Pharmacologically evoked fictive motor patterns in the acutely spinalized marmoset monkey (*Callithrix jacchus*). *Experimental Brain Research, 122,* 351-361.

Ferguson, B., Matyszak, M. K., Esiri, M. M., & Perry, V. H. (1997). Axonal damage in acute multiple sclerosis lesions, *Brain, 120,* 393-399.

Fischer, J. S., Rudick, R. A., Cutter, S. C., & Reingold, S. C. (1999). The Multiple Sclerosis Functional Composite measure (MSFC): An integrated approach to MS clinical outcome assessment. *Multiple Sclerosis, 5,* 244-250.

Garcia-Rill, E. (1986). The basal ganglia and locomotor regions. *Brain Research Reviews, 11,* 47-63.

Gefen, A., Megido-Ravid, M., & Itchak, Y. (2001). In vivo biomechanical behavior of the human heel pad during the stance phase of gait. *Journal of Biomechanics, 34,* 1661-1665.

Gehlsen, G., Beekman, K., Assmann, N., Winant, D., Seidle, M., & Carter, A. (1986). Gait characteristics in multiple sclerosis: Progressive changes and effects of exercise on parameters. *Archives of Physical Medicine and Rehabilitation, 67,* 536-539.

Givon, U., Zeilig, G., & Achiron, A. (2009). Gait analysis in multiple sclerosis: Characterization of temporal-spatial parameters using GAITRite functional ambulation system. *Gait & Posture, 29,* 138-142.

Graham Brown, T. (1911). The intrinsic factors in the act of progression in the mammal. *Proceeding of the Royal Society of London, 84B,* 308-319.

Grillner, S., Wallén, P., Saitoh, K., Kozlov, A., & Robertson, B. (2008). Neural bases of goal-directed locomotion in vertebrates - an overview. *Brain Research Reviews, 57,* 2-12.

Gutierrez, G. M., Chow, J. W., Tillman, M. D., McCoy, S. C., Castellano, V., & White, L. J. (2005). Resistance training improves gait kinematics in persons with multiple sclerosis. *Archives of Physical Medicine and Rehabilitation, 86,* 1824-1829.

Hagemeister, N., Parent, G., Van de Putte, M., St-Onge, N., Duval, N., & de Guise, J. (2005). A reproducible method for studying three-dimensional knee kinematics. *Journal of Biomechanics, 38,* 1926-1931.

Harvill, L. M. (1991). Standard error of measurement. *Educational Measurement: Issues and Practice, 10,* 33-41.

Heesen, C., Böhm, J., Reich, C., Kasper, J., Goebel, M., & Gold, S. M. (2008). Patient perception of bodily functions in multiple sclerosis: gait and visual function are the most valuable. *Multiple Sclerosis, 14,* 988-991.

Holden, M. K., Gill, K. M., & Magliozzi, M. R. (1986). Gait assessment for neurologically impaired patients: Standards for outcome assessment. *Physical Therapy, 66,* 1530-1539.

Howell, D. C. (2007). *Statistical methods in psychology* (6th ed.). Belmont, CA: Thomson Wadsworth.

Hultborn, H. & Nielsen, J. B. (2007). Spinal control of locomotion – from cat to man. *Acta Physiologica, 189,* 111-121.

Inman, V., Ralston, H., & Todd, F. (1981). *Human walking.* Baltimore, MD: Williams and Wlikins.

Kawamura, C. M., de Morais Filho, M. C., Barreto, M. M., de Paula Asa, S. K., Juliano, Y., & Novo, N. F. (2007). Comparison between visual and three-dimensional gait analysis in patients with spastic diplegic cerebral palsy. *Gait & Posture, 25,* 18-24.

Kirtley, C., Whittle, M. W., & Jefferson, R. J. (1985). Influence of walking speed on gait parameters. *Journal of Biomedical Engineering, 7,* 282-288.

Krebs, D. E., Edelstein, J. E., & Fishman, S. (1985). Reliability of observational kinematic gait analysis. *Physical Therapy, 65,* 1027-1033.

Kurtzke, J. F. (1983). Rating neurologic impairment in multiple sclerosis: An expanded disability status scale (EDSS). *Neurology, 33,* 1444-1452.

Lance, J. W. (1980). Symposium synopsis. In Feldman, R. G., Young, R. R., & Koella, W. P. (Eds.), *Spasticity, disordered motor control* (pp. 485-500). Miami, FL: Symposia Specialists.

Lee, D. N. (1980). The optic flow field: The foundation of vision. *Philosophical Transactions of the Royal Society of London, 290,* 169-178.

Lublin, F. D. & Reingold, S. C. (1996). Defining the clinical course of multiple sclerosis: Results of an international survey. *Neurology, 46,* 907-911.

Martin, C. L., Phillips, B. A., Kilpatrick, T. J., Butzkueven, H., Tubridy, N., McDonald, E. et al. (2006). Gait and balance impairment in early multiple sclerosis in the absence of clinical disability. *Multiple Sclerosis, 12,* 620-628.

Maynard, V., Bakheit, A. M. O., Oldham, J., & Freeman, J. (2003). Intra-rater and inter-rater reliability of gait measurements with CODA mpx30 motion analysis system. *Gait and Posture, 17,* 59-67.

McDonald, W. I. (1974). Pathophysiology in multiple sclerosis. *Brain, 97,* 179-196.

McDonald, W. I., Compston, A., Edan, G., Goodkin, D., Hartung, H., Lublin, F. D., et al. (2001). Recommended diagnostic criteria for multiple sclerosis: Guidelines from the International Panel on the diagnosis of multiple sclerosis. *Annals of Neurology, 50,* 121-127.

McDonough, A. L., Batavia, M., Chen, F. C., Kwon, S., & Ziai, J. (2001). The validity and reliability of the GAITRite system's measurements: A preliminary evaluation. *Archives of Physical Medicine & Rehabilitation, 82,* 419-425.

McGinley, J. L., Baker, R., Wolfe, R., & Morris, M. E. (2009). The reliability of three-dimensional kinematic gait measurements: A systematic review. *Gait & Posture, 29,* 360-369.

Menz, H. B., Latt, M. D., Tiedemann, A., Kwan, M. M. S., & Lord, S. (2004). Reliability of the GAITRite® walkway system for the quantification of temporo-spatial parameters of gait in young and older people. *Gait and Posture, 20,* 20-25.

Morales, Y., Parisi, J.E, & Lucchinetti, C. F. (2006). The pathology of multiple sclerosis: Evidence for heterogeneity. *Advances in Neurology, 98,* 27-46.

Moriguchi, C. S., Sato, T. O., & Coury, H. J. C. G. (2007). Ankle movements during normal gait evaluated by flexible electrogoniometer. *Revista Brasileira de Fisioterapia, 11,* 205-211.

Morita, H., Crone, C., Christnehuis, D., Petersen, T., & Nielsen, J. B. (2001). Modulation of presynaptic inhibition and disynaptic reciprocal Ia inhibition during voluntary movement in spasticity. *Brain, 124,* 826-837.

Morris, M. E., Cantwell, C., Vowels, L., & Dodd, K. (2002). Changes in gait and fatigue from morning to afternoon in people with multiple sclerosis. *Journal of Neurology, Neurosurgery, & Psychiatry, 72,* 361-365.

Nelson, A. J., Zwick, D., Brody, S., Doran, C., Pulver, L., Rooz, G., et al. (2002). The validity of the GaitRite and the Functional Ambulation Performance scoring system in the analysis of Parkinson gait. *NeuroRehabilitation, 17,* 255-262.

Ng, A. V., Miller, R. G., Gelinas, D., & Kent-Braun, J. A. (2004). Functional relationships of central and peripheral muscle alterations in multiple sclerosis. *Muscle & Nerve, 29,* 843-852.

Noseworthy, J. H., Lucchinetti, C., Rodriquez, M., & Weinshenker, B. G. (2000). Multiple sclerosis. *The New England Journal of Medicine, 343,* 938-952.

O'Brien, J. F., Bodenheimer, R. E., Brostow, G. J., & Hodgins, J. K. (2000). Automatic joint parameter estimation from magnetic motion capture data. In *Proceedings of Graphics Interface 2000,* pp. 53-60.

Olgiati, R., Burgunder, J. M., & Mumenthaler, M. (1988). Increased energy cost of walking in multiple sclerosis: Effect of spasticity, ataxia, and weakness. *Archives of Physical Medicine and Rehabilitation, 69,* 846-849.

Orsnes, G. B., Sorensen, P. S., Larsen, T. K., & Ravnborg, M. (2000). Effect of baclofen on gait in spastic MS patients. *Acta Neurologica Scandinavica, 101,* 244-248.

Paltamaa, J., Sarasoja, T., Wikström, J., & Mälkiä, E. (2006). Physical functioning in multiple sclerosis: A population-based study in Central Finland. *Journal of Rehabilitation Medicine, 38,* 339-345.

Peters, A., Galna, B., Sangeux, M., Morris, M., & Baker, R. (2010). Quantification of soft tissue artifact in lower limb human motion analysis: A systematic review. *Gait & Posture, 31,* 1-8.

Piriyaprasarth, P., Morris, M. E., Winter, A., & Bialocerkowski, A. E. (2008). The reliability of knee joint position testing using electrogoniometry. *BMC Musculoskeletal Disorders, 9:*6.

Polman, C. H., Reingold, S. C., Edan, G., Filippi, M., Hartung, H., Kappos, L., et al. (2005). Diagnostic criteria for multiple sclerosis: 2005 revisions to the "McDonald Criteria." *Annals of Neurology, 58,* 840-846.

Poser, C. M., Paty, D. W., Scheinberg, L., McDonald, W. I., Davis, F. A., Ebers, G. C., et al. (1983). New diagnostic criteria for multiple sclerosis: Guidelines for research protocols. *Annals of Neurology, 13,* 227-231.

Pryse-Phillips, W. & Sloka, S. (2006). Etiopathogenesis and epidemiology: Clues to etiology. In S. D. Cook (Ed.), *Handbook of multiple sclerosis* (pp. 1-40). New York, NY: Taylor & Francis Group.

Ralston, H. J. (1958). Energy-speed relation and optimal speed during level walking. *European Journal of Applied Physiology and Occupational Physiology, 17,* 277-283.

Rinker II, J. R., Naismith, R. T., & Cross, A. H. (2006). Multiple sclerosis: An autoimmune disease of the central nervous system? In S. D. Cook (Ed.), *Handbook of multiple sclerosis* (pp. 95-112). New York, NY: Taylor & Francis Group.

Rizzo, M. A., Hadjimichael, O. C., Preiningerova, J., & Vollmer, T. L. (2004). Prevalence and treatment of spasticity reported by multiple sclerosis patients. *Multiple Sclerosis, 10,* 589-595.

Rome, K. & Cowieson, F. (1996). A reliability study of the universal goniometer, fluid goniometer, and electrogoniometer for the measure of ankle dorsiflexion. *Foot & Ankle International, 17,* 28-32.

Rossignol, S., Dubuc, R., & Gossard, J. P. (2006). Dynamic sensorimotor interactions in locomotion. *Physiological Reviews, 86,* 89-154.

Rowe, P. J., Myles, C. M., Hillmann, S. J., & Hazlewood, M. E. (2001). Validation of flexible electrogoniometry as a measure of joint kinematics. *Physiotherapy, 87,* 479-488.

Rudick, R., Antel, J., Confavreaux, C., Cutter, G., Ellison, G., Fischer, J., et al. (1996). Clinical outcomes assessment in multiple sclerosis. *Annals of Neurology, 40,* 469-479.

Rudick, R. A., Cutter, G., Baier, M., Fisher, E., Dougherty, D., Weinstock-Guttman, B., et al. (2001). Use of the Multiple Sclerosis Functional Composite to predict disability in relapsing MS. *Neurology, 56,* 1324-1330.

Saleh, M. & Murdoch, G. (1985). In defence of gait analysis: Observation and measurement in gait assessment. *The Journal of Bone and Joint Surgery, 67 B,* 237-241.

Saunders, J. B., Inman, V. T., & Eberhart, H. D. (1953). The major determinants in normal and pathological gait. *The Journal of Bone and Joint Surgery, 35,* 543-558.

Scheinberg, L., Holland, N., Larocca, N., Laitin, P., Bennett, A., & Hall, H. (1980). Multiple sclerosis: Earning a living. *New York State Journal of Medicine, 80,* 1395-1400.

Schwartz, J. H. & Westbrook, G. L. (2000). The cytology of neurons. In E. R. Kandell, J. H. Schwartz, & T. M. Jessell (Eds.), *Principles of neural science* (4th ed., pp. 67-104). New York: McGraw-Hill.

Sherington, C. S. (1906). On the proprio-ceptive system, especially in its reflex aspect. *Brain, 29,* 467-482.

Shrout, P. E. & Fleiss, J. L. (1979). Intraclass correlations: Uses in assessing rater reliability. *Psychological Bulletin, 86,* 420-428.

Sinkjaer, T., Andersen, J. B., & Nielsen, J. F. (1996). Impaired stretch reflex and joint torque modulation during spastic gait in multiple sclerosis patients. *Journal of Neurology, 243,* 566-574.

Solaro, C., Brichetto, G., Casadio, M., Roccatagliata, L., Ruggiu, P., Mancardi, G. L., et al. (2007). Subtle upper limb impairment in asymptomatic multiple sclerosis subjects. *Multiple Sclerosis, 13,* 428-432.

Sosnoff, J. J., Shin, S., & Motl, R. W. (2010). Multiple sclerosis and postural control: The role of spasticity. *Archives of Physical Medicine and Rehabilitation, 91,* 93-99.

Stagni, R., Fantozzi, S., Cappello, A., & Leardini, A. (2005). Quantification of soft tissue artefact in motion analysis by combining 3D fluoroscopy and sterephotogrammetry: a study on two subjects. *Clinical Biomechanics, 20,* 320-329.

Stoquart, G., Detrembleur, C., & Lejeune, T. (2008). Effect of speed on kinematic, kinetic, electromyographic and energetic reference values during treadmill walking. *Clinical Neurophysiology, 38,* 105-116.

Sutliff, M. H. (2010). Contribution of impaired mobility to patient burden in multiple sclerosis. *Current Medical Research & Opinion, 26,* 109-119.

Takakusaki, K., Habaguchi, T., Ohtinata-Sugimoto, J., Saitoh, K., & Sakamoto, T. (2003). Baal ganglia efferents to the brainstem centers controlling postural muscle tone and locomotion: A new concept for understanding motor disorders in basal ganglia dysfunction. *Neuroscience, 119,* 293-308.

Thomas, J. R., Nelson, J. K., & Silverman, S. J. (2005). *Research methods in physical activity* (5th ed.). Champaign, IL: Human Kinetics.

Thoumie, P., Lamotte, D., Cantalloube, S., Faucher, M., & Amarenco, G. (2005). Motor determinants of gait in 100 ambulatory patients with multiple sclerosis. *Multiple Sclerosis, 11,* 485-491.

van der Linden, M. L., Rowe, P. J., & Nutton, R. W. (2008). Between-day repeatability of knee kinematics during functional tasks recorded using flexible electrogoniometry. *Gait & Posture, 28,* 292-296.

van Uden, C. J. T. & Besser, M. (2004). Test-retest reliability of temporal and spatial gait characteristics measured with an instrumented walkway system (GAITRite®). *BMC Musculosketal Disorders, 5*:13.

Vincent, W. J. (2005). *Statistics in kinesiology* (3rd ed.). Champaign, IL: Human Kinetics.

Webster, K. E., Wittwer, J., & Feller, J. A. (2005). Validity of the GAITRite® walkway system for the measurement of averaged and individual step parameters of gait. *Gait and Posture, 22,* 317-321.

Weinshenker, B. G., Bass, B., Rice, G. P. A., Noseworthy, J., Carriere, W., Baskerville, J., et al. (1989). The natural history of multiple sclerosis: A geographically based study: I. Clinical course and disability. *Brain, 112,* 133-146.

Weir, J. P. (2005). Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *Journal of Strength and Conditioning Research, 19,* 231-240.

Whittle, M. W. (1996). Clinical gait analysis: A review. *Human Movement Science, 15,* 369-387.

Williams, G., Morris, M. E., Schache, A., & McRory, P. (2009). Observational gait analysis in traumatic brain injury: Accuracy of clinical judgment. *Gait & Posture, 29,* 454-459.

Winter, D. A. (1991). *The biomechanics and motor control of human gait: Normal, elderly and pathological* (2nd ed.). Waterloo, ON: University of Waterloo.

Winter, D. A. (2005). *Biomechanics and motor control of human movement* (3rd ed.). Hoboken, NJ: John Wiley & Sons.

Yack, H. J. (1984). Techniques for clinical assessment of human movement. *Physical Therapy, 64,* 1821-1830.

Appendix A - Study 1 University of Saskatchewan
Behavioural Research Ethics Board Approval

UNIVERSITY OF SASKATCHEWAN

Behavioural Research Ethics Board (Beh-REB)

Certificate of Re-Approval

PRINCIPAL INVESTIGATOR	DEPARTMENT	Beh #
Joel Lanovaz	Kinesiology	08-120

INSTITUTION (S) WHERE RESEARCH WILL BE CARRIED OUT

University of Saskatchewan
Saskatoon SK

STUDENT RESEARCHER(S)

Kristopher Beyer

SPONSORING AGENCIES

NATURAL SCIENCES & ENGINEERING RESEARCH COUNCIL OF CANADA (NSERC) DI

TITLE

Computational Modeling of the Musculoskeletal System

RE-APPROVED ON	EXPIRY DATE
19-May-2009	18-May-2010

Full Board Meeting

Delegated Review ✔

CERTIFICATION

The University of Saskatchewan Behavioural Research Ethics Board has reviewed the above-named research project. The proposal was found to be acceptable on ethical grounds. The principal investigator has the responsibility for any other administrative or regulatory approvals that may pertain to this research project, and for ensuring that the authorized research is carried out according to the conditions outlined in the original protocol submitted for ethics review. This Certificate of Approval is valid for the above time period provided there is no change in experimental protocol or consent process or documents.

Any significant changes to your proposed method, or your consent and recruitment procedures should be reported to the Chair for Research Ethics Board consideration in advance of its implementation.

ONGOING REVIEW REQUIREMENTS

In order to receive annual renewal, a status report must be submitted to the REB Chair for Board consideration within one month of the current expiry date each year the study remains open, and upon study completion. Please refer to the following website for further instructions: http://www.usask.ca/research/ethics_review/

John Rigby

John Rigby, Chair
University of Saskatchewan
Behavioural Research Ethics Board

Please send all correspondence to

Research Ethics Office
University of Saskatchewan
Box 5000 RPO University
1607-110 Gymnasium Place
Saskatoon SK Canada S7N 4J8

Appendix B - Study 1 Participant Information and Consent Form

Research Participant Information and Consent Form

Title: Validation of knee and ankle flexible electrogoniometry during gait

Name of Researcher:

Principal Investigator:
Joel Lanovaz, Ph.D., Assistant Professor
College of Kinesiology
87 Campus Drive, University of Saskatchewan
Phone: (306)-966-1073
Email: joel.lanovaz@usask.ca

Student Researcher:
Kristopher Beyer, M Sc. Candidate
College of Kinesiology
87 Campus Drive, University of Saskatchewan
Email: kit.beyer@usask.ca

Purpose: You are being invited to participate in a research study because we want to explore the ability of flexible electrogoniometry to reliably measure joint angles during gait (walking). A flexible electrogoniometer is a small flexible electrical device that is placed across the joint to measure the angle of that joint. The data collected from this device will be compared with data from a camera system that is known to be able to measure joint angles during gait.

Possible benefits: There are no anticipated benefits from this study to you directly, but if flexible electrogoniometry is found reliable it may provide a relatively simple, portable and inexpensive tool for assessing and studying gait in both healthy and unhealthy populations.

Procedures: If you agree to participate in this study, the following will happen:

Your height and weight will be measured and recorded and you will be asked to change into athletic shorts (change rooms are across the hall). A set of approximately 30 small reflective spheres will be attached to various locations on your legs and torso. Some of the spheres will be grouped together on a lightweight plastic plate and attached to your leg using an elastic strap. Other spheres will be attached directly to your skin using two-sided medical tape.

Electrogoniometers will be attached across the outside of your ankles and knees by fastening the end-blocks above and below each joint using double-sided medical tape. Small flat force switches will be placed on the heel and great toe of each of your feet to indicate when your heels and toes make contact with the ground. Wires will connect the electrogoniometers and force switches to the data logger which will be attached to a belt around your waist. The data logger transmits data wirelessly to a computer so there are no cables to impede your walking. The entire system is lightweight (approx. 400g) to ensure minimal discomfort.

You will then be asked to walk at a natural pace along a runway in the lab. A series of special cameras will follow your motion during this time by tracking the locations of the spheres. There is also a special platform imbedded into the floor of the walkway which will measure the forces you apply on the ground as you walk over it. You will be asked to walk in both directions 10 to 20 times at both a comfortable self-selected pace and a comfortable brisk pace.

You will be asked to visit the lab for two sessions within 48 hours of each other. In each session you will complete the entire procedure twice, once each with two different testers, for a time commitment of approximately 60 to 90 minutes per session.

Foreseeable risks, side effects or discomfort: The risks from this study are minimal and are no more than what you would have in normal everyday activity. The movements that you will be performing do not require much physical exertion. However, if you feel tired or uncomfortable, you may ask to rest at any time and for as long as you need. There may be some discomfort on your skin from the adhesive tape that temporarily sticks the spheres, electrogoniometers, and force switches to your skin, but this is rare. There may also be unforeseen and unknown risks during the study, or after the study has been completed.

Voluntary Participation: Before you decide, it is important for you to understand what the research involves. This consent form will tell you about the study, why the research is being done, what will happen to you during the study and the possible benefits, risks and discomforts.

If you wish to participate, you will be asked to sign this form. Your participation is entirely voluntary, so it is up to you to decide whether or not to take part in this study. If you do decide to take part in this study, you are free to withdraw at any time without giving any reasons for your decision and your refusal to participate will not affect your relationship with any of the researchers or the University of Saskatchewan, and will not affect your academic standing if you are a student at the university. Please take time to read the following information carefully before you decide.

Storage of Data: Computer based data are stored on password protected digital media (i.e. DVD) in a locked cabinet in the Musculoskeletal Biomechanics lab to which only the investigators will have access. Digital video files are also stored in a locked cabinet in the lab, with access restricted to the investigators. The data will be used for dissertation and publication purposes only, and will be retained for a minimum of five years post-publication. Normally data is retained for a period of five years post-publication, after which time it may be destroyed.

Confidentiality: All information and data collected are coded to maintain confidentiality. To ensure confidentiality, your data will be assigned a numerical identification code. Your name will only be used for contact purposes regarding feedback and your name will not be associated with any data. Publication of data will generally be reported as group values, however, in the event that individual's data is reported, participant codes will be referenced. Although digital cameras are used extensively during data collection, image records will not be used to identify any specific participant. Further, all video data will only be accessible to the researchers (see the section on Storage of Data).

Dissemination of Results: The data from this will be presented by the researchers at academic conferences and published in peer-reviewed academic journals. If you wish to receive a lay person's summary of the results of this study after it is complete, please contact Dr. Joel Lanovaz by phone (306-966-1073) or e-mail (joel.lanovaz@usask.ca). This summary will be an aggregate of all results and not your individual results.

Appendix B - Study 1 Participant Information and Consent Form

Voluntary Withdrawal: Your participation in this research is entirely voluntary. You may withdraw from this study at any time. If you decide to enter the study and to withdraw at any time in the future, there will be no penalty or loss of benefits to which you are otherwise entitled. If you choose to enter the study and then decide to withdraw at a later time, all data collected about you during your enrolment in the study will be retained for analysis.

Questions: If you have any questions concerning the study, please feel free to ask them at any point; you are also free to contact the researchers at Dr. Joel Lanovaz at 306-966-1073 (collect calls accepted) or by e-mail provided if you have any questions at a later time. This research project was reviewed and approved on ethical grounds by the University of Saskatchewan Behavioural Research Ethics Board.

If you have questions about your rights as a research subject, you should contact the Chair of the Behavioural Research Ethics Board, University of Saskatchewan at (306) 966-2084. Again, this number can be called collect if you are phoning long distance.

By signing below, I confirm the following:

- I have read or have had this read to me and understood the research subject information and consent form.
- I have had sufficient time to consider the information provided and to ask for advice if necessary.
- I have had the opportunity to ask questions and have had satisfactory responses to my questions.
- I understand that all of the information collected will be kept confidential and that the result will only be used for scientific objectives.
- I understand that my participation in this study is voluntary and that I am completely free to refuse to participate or to withdraw from this study at any time without changing in any way affect my academic standing or my relationship with members of the research team.
- I understand that I am not waiving any of my legal rights as a result of signing this consent form.
- I understand that there is no guarantee that this study will provide any benefits to me.
- I have read this form and I freely consent to participate in this study.
- I have been told that I will receive a dated and signed copy of this form

Participant's Signature:_____ Date: _____

Individual conducting the consent process:_____

Date: _____

Appendix C - Study 2 University of Saskatchewan
Behavioural Research Ethics Board Approval

UNIVERSITY OF SASKATCHEWAN

Behavioural Research Ethics Board (Beh-REB)

Certificate of Approval

PRINCIPAL INVESTIGATOR	DEPARTMENT	BEH#
Lawrence Brawley	Kinesiology	09-003

INSTITUTION(S) WHERE RESEARCH WILL BE CONDUCTED
University of Saskatchewan

SUB-INVESTIGATOR(S)
Joel Lanovaz

STUDENT RESEARCHERS
Kristopher Beyer

SPONSOR
NATURAL SCIENCES & ENGINEERING RESEARCH COUNCIL OF CANADA (NSERC)
CANADA FOUNDATION FOR INNOVATION
CAMECO CORPORATION
Canada Research Chair (CRC) funding Tier 1

TITLE
Kinematic gait abnormalities in people with multiple sclerosis

ORIGINAL REVIEW DATE	APPROVAL ON	APPROVAL OF:	EXPIRY DATE
10-Jan-2009	30-Jan-2009	Ethics Application Consent Protocol	29-Jan-2010

Full Board Meeting ☐ Date of Full Board Meeting:

Delegated Review ☒

CERTIFICATION
The University of Saskatchewan Behavioural Research Ethics Board has reviewed the above-named research project. The proposal was found to be acceptable on ethical grounds. The principal investigator has the responsibility for any other administrative or regulatory approvals that may pertain to this research project, and for ensuring that the authorized research is carried out according to the conditions outlined in the original protocol submitted for ethics review. This Certificate of Approval is valid for the above time period provided there is no change in experimental protocol or consent process or documents.

Any significant changes to your proposed method, or your consent and recruitment procedures should be reported to the Chair for Research Ethics Board consideration in advance of its implementation.

ONGOING REVIEW REQUIREMENTS
In order to receive annual renewal, a status report must be submitted to the REB Chair for Board consideration within one month of the current expiry date each year the study remains open, and upon study completion. Please refer to the following website for further instructions: *http://www.usask.ca/research/ethics_review/*

John Rigby, Chair

Please send all correspondence to:

Research Ethics Office
University of Saskatchewan
Box 5000 RPO University, 1602-110 Gymnasium Place
Saskatoon SK S7N 4J8

Appendix C - Study 2 University of Saskatchewan
Behavioural Research Ethics Board Approval

UNIVERSITY OF SASKATCHEWAN

Behavioural Research Ethics Board (Beh-REB)

Certificate of Approval
Study Amendment

PRINCIPAL INVESTIGATOR	DEPARTMENT	Beh #
Lawrence Brawley	Kinesiology	09-03

INSTITUTION(S) WHERE RESEARCH WILL BE CARRIED OUT
University of Saskatchewan

SUB-INVESTIGATOR(S)
Joel Lanovaz

STUDENT RESEARCHER(S)
Kristopher Beyer

SPONSORING AGENCIES
NATURAL SCIENCES & ENGINEERING RESEARCH COUNCIL OF CANADA (NSERC)
CANADA FOUNDATION FOR INNOVATION
CAMECO CORPORATION
Canada Research Chair (CRC) funding Tier 1

TITLE
Kinematic gait abnormalities in people with multiple sclerosis

APPROVAL OF	APPROVED ON	CURRENT EXPIRY DATE
Revised recruitment protocol	05-Feb-2009	29-Jan-2010

Full Board Meeting ☐ Date of Full Board Meeting:

Delegated Review ☒

CERTIFICATION
The University of Saskatchewan Behavioural Research Ethics Board has reviewed the above-named research project. The proposal was found to be acceptable on ethical grounds. The principal investigator has the responsibility for any other administrative or regulatory approvals that may pertain to this research project, and for ensuring that the authorized research is carried out according to the conditions outlined in the original protocol submitted for ethics review. This Certificate of Approval is valid for the above time period provided there is no change in experimental protocol or consent process or documents.

Any significant changes to your proposed method, or your consent and recruitment procedures should be reported to the Chair for Research Ethics Board consideration in advance of its implementation.

ONGOING REVIEW REQUIREMENTS
In order to receive annual renewal, a status report must be submitted to the REB Chair for Board consideration within one month of the current expiry date each year the study remains open, and upon study completion. Please refer to the following website for further instructions: *http://www.usask.ca/research/ethics_review/*

John Rigby

John Rigby, Chair
University of Saskatchewan
Behavioural Research Ethics Board

Please send all correspondence to:

Research Ethics Office
University of Saskatchewan
Box 5000 RPO University, 1602-110 Gymnasium Place
Saskatoon SK S7N 4J8

Multiple Sclerosis Gait Characteristics Study

Screening Interview

Participant ID:	Disease Status: No CIS MS	
Visit Date: DD/MM/YYYY	Birthdate: DD/MM/YYYY	
Gender: F M	Height: cm	Weight: kg
Dominant Hand: L R	Leg length (GT): L: cm R: cm	

Medical History

Neurological conditions other than MS.
Musculoskeletal problems (arthritis, gout, injury).
Cardiovascular disease.
Other conditions causing disability of lower limbs.

Surgical History

Type	Date	Reason	Complications

Appendix D - Study 2 Screening Interview

Family Medical History

Any family members with MS or have a clinical attack suggestive of MS.
Any family members with other significant conditions (Parkinson's, Alzheimer's).

Multiple Sclerosis-specific Medical History

Significant childhood illnesses
Age at MS onset or first clinical attack suggestive of MS onset.
Age at diagnosis of MS. Diagnosing physician.
Last clinical assessment. Symptom change since last clinical assessment.

Appendix E – Expanded Disability Status Scale (EDSS)

From Kurtzke, 1983.

Functional Systems.

Pyramidal Functions

0. Normal
1. Abnormal signs without disability.
2. Minimal disability.
3. Mild of moderate paraparesis or hemiparesis; severe monoparesis.
4. Marked paraparesis or hemiparesis; moderate quadriparesis; or monoplegia.
5. Paraplegia, hemiplegia, or marked quadriparesis.
6. Quadriplegia.

V. Unknown.

Cerebellar Functions

0. Normal.
1. Abnormal signs without disability.
2. Mild ataxia.
3. Moderate truncal or limb ataxia.
4. Severe ataxia, all limbs.
5. Unable to perform coordinated movements due to ataxia.

V. Unknown.

X. Is used throughout after each number when weakness (grade 3 or more on pyramidal) interferes with testing.

Brain Stem Functions

0. Normal.
1. Signs only.
2. Moderate nystagmus or other mild disability.
3. Severe nystagmus, marked extraocular weakness, or moderate disability of other cranial nerves.
4. Marked dysarthria or other marked disability.
5. Inability to swallow or speak.

V. Unknown.

Sensory Functions

0. Normal.
1. Vibration or figure-writing decrease only, in one or two limbs.
2. Mild decrease in touch or pain or position sense, and/or moderate decrease in vibration in one or two limbs; or vibratory (c/s figure writing) decrease alone in three or four limbs.

3. Moderate decrease in touch or pain or position sense, and/or essentially lost vibration in one or two limbs; or mild decrease in touch or pain and/or moderate decrease in all proprioceptive tests in three or four limbs.
4. Marked decrease in touch or pain or loss of proprioception, alone or combined, in one or two limbs; or moderate decrease in touch or pain and/or severe proprioceptive decrease in more than two limbs.
5. Loss (essentially) of sensation in one or two limbs; or moderate decrease in touch or pain and/or loss of proprioception for most of the body below the head.
6. Sensation essentially lost below the head.

V. Unknown.

Bowel and Bladder Functions

0. Normal.
1. Mild urinary hesitancy, urgency, or retention.
2. Moderate hesitancy, urgency, retention of bowel or bladder, or rare urinary incontinence.
3. Frequent urinary incontinence.
4. In need of almost constant catheterization.
5. Loss of bladder function.
6. Loss of bladder and bowel function.

V. Unknown.

Visual (or Optic) Functions

0. Normal.
1. Scotoma with visual acuity (corrected) better than 20/30.
2. Worse eye with scotoma with maximal visual acuity (corrected) of 20/30 to 20/59.
3. Worse eye with large scotoma, or moderate decrese in fields, but with maximal visual acuity (corrected) of 20/60 to 20/99.
4. Worse eye with marked decrease of fields and maximal visual acuity (corrected) of 20/100 to 20/200; grade 3 plus maximal acuity of better eye of 20/60 or less.
5. Worse eye with maximal visual acuity (corrected) less than 20/200; grade 4 plus maximal acuity of better eye of 20/60 or less.
6. Grade 5 plus maximal visual acuity of better eye of 20/60 or less.

V. Unknown.

X. Is added to grades 0 to 6 for presence of temporal pallor.

Cerebral (or Mental) Functions

0. Normal.
1. Mood alteration only (Does not affect DDS score).
2. Mild decrease in mentation.
3. Moderate decrease in mentation.

4. Marked decrease in mentation (chronic brain syndrome–moderate).
5. Dementia or chronic brain syndrome–severe or incompetent.

V. Unknown.

Other Functions

0. None.
1. Any other neurologic findings attributed to MS (specify).

V. Unknown.

Expanded Disability Status Scale (EDSS)

0 = Normal neurologic exam (all grade 0 in Functional Systems [FS]; Cerebral grade 1 acceptable).

1.0 = No disability, minimal signs in one FS (i.e,, grade 1 excluding Cerebral grade 1).

1.5 = No disability signs in more than one FS (more than one grade 1 excluding Cerebral grade 1).

2.0 = Minimal disability in one FS (one FS grade 2, others 0 or 1).

2.5 = Minimal disability in two FS (two FS grade 2, others 0 or 1).

3.0 = Moderate disability in one FS (one FS grade 3, others 0 or 1).

3.5 = Fully ambulatory but with moderate disability in one FS (one grade 3) and one or two FS grade 2; or two FS grade 3; or five FS grade 2 (others 0 or 1).

4.0 = Fully ambulatory without aid, self-sufficient, up and about some 12 hours a day despite relatively severe disability consisting of one FS grade 4 (others 0 or 1), or combination of lesser grades exceeding limits of previous steps. Able to walk without aid or rest some 500 meters.

4.5 = Fully ambulatory without aid, up and about much of the day, able to work a full day, may otherwise have some limitation of full activity or require minimal assistancel characterized by relatively severe disability, usually consisting of one FS grade 4 (others 0 or 1) or combinations of lesser grades exceeding limits of previous steps. Able to walk without aid or rest for some 300 meters.

Appendix E – Expanded Disability Status Scale (EDSS)

5.0 = Ambulatory without aid or rest for about 200 meters; disability severe enough to impair full daily activities (eg, to work full day without special provisions). (Usual FS equivalents are one grade 5 alone, others 0 or 1; or combinations of lesser grades usually exceeding specifications for step 4.0.)

5.5 = Ambulatory without aid or rest for about 100 meters; disability severe enough to preclude full daily activities. (Usual FS equivalents are one grade 5 alone, others 0 or 1l or combinations of lesser grades usually exceeding those for step 4.0.)

6.0 = Intermittent or unilateral constant assistance (cane, crutch, or brace) required to walk about 100 meters with or without resting. (Usual FS equivalents are combinations with more than two FS grade 3+.)

6.5 = Constant bilateral assistance (canes, crutches, or braces) required to walk about 20 meters without resting. (Usual FS equivalents are combinations with more than two FS grade 3+.)

7.0 = Unable to walk beyond about 5 meters even with aid, essentially restricted to wheelchair; wheels self in standard wheelchair and transfers alone; up and about in w/c some 12 hours a day. (Usual FS equivalents are combinations with more than one FS grade 4+; very rarely, pyramidal grade 5 alone.)

7.5 = Unable to take more than a few steps; restricted to wheelchair; may need aid in transfer; wheels self but cannot carry on in standard wheelchair a full day; may require motorized wheelchair. (usual FS equivalents are combinations with more than one FS grade 4+.)

8.0 = Essentially restricted to bed or chair or perambulated in wheelchair, but may be out of bed itself much of the day; retains many self-care functions; generally has effective use of arms. (Usual FS equivalents are combinations, generally grade 4+ in several symptoms.)

8.5 = Essentially restricted to bed much of the day; has some effective use of arm(s); retains some self-care functions. (Usual FS equivalents are combinations, generally 4+ in several systems.)

9.0 = Helpless bed patient; can communicate and eat. (Usual FS equivalents are combinations, mostly grade 4+.)

Appendix E – Expanded Disability Status Scale (EDSS)

9.5 = Totally helpless bed patient; unable to communicate effectively or eat/swallow. (Usual FS equivalents are combination, almost all grade 4+.)

10. = Death due to MS.

Appendix F - Study 2 Research Participant Information and Consent Form

RESEARCH PARTICIPANT INFORMATION AND CONSENT FORM

UNIVERSITY OF SASKATCHEWAN

You are invited to participate in a research study entitled 'Kinematic gait abnormalities in people with multiple sclerosis.' Please read this form carefully, and feel free to ask questions you might have.

RESEARCHERS

Larry Brawley, Ph.D., Canada Research Chair,
College of Kinesiology,
University of Saskatchewan,
Ph: 306-966-1076

Kit Beyer, M.Sc. Candidate,
College of Kinesiology,
University of Saskatchewan,
Ph: 306-966-1123

PURPOSE AND PROCEDURES

The purpose of this study is to characterize walking abnormalities among individuals with MS by comparing the walking characteristics of people with MS to a group of people without MS using simple, portable and inexpensive measurement tools. Your total time commitment will be one 60-90 minute session.

First, you will be asked a few questions about yourself and the status of your MS. You are free to answer only those questions that you are comfortable with.

Next, your walking will be assessed while you walk 25 feet three times at a self-selected comfortable pace and three more times at a brisk but comfortable pace. You will be allowed a seated rest in between sets and any other time you feel you need it. The devices used to assess your walking are described below. Please ask any questions you have about each device and what it measures if you are still unclear after reading the descriptions.

A 10m mat will be placed on the 25-foot course. You will simply walk over this mat and it will collect data about the timing and placement of your footsteps. A number of small flexible devices will be placed across both of your knee and ankle joints to measure the angles of these joints as you walk. These devices will be adhered to the skin with double-sided medical tape and connected by wires to a small data pack attached to a belt around your waist.

After your walking is assessed, you will be asked to complete the Multiple Sclerosis Functional Composite, which consists of three functional tests measuring arm, leg and cognitive function. The Timed 25-Foot Walk measures arm function by measuring the time taken to travel a clearly marked 25-foot path as quickly, but safely, as possible. The 9-Hole Peg Test measures arm function by measuring the time taken to individually put nine pegs into a pegboard and then remove them. The Paced Auditory Serial Addition Test measures cognitive function by tracking

errors during an addition task where a digit is verbally presented every three seconds and must be added to the previous digit.

Finally, we also request your permission to contact the physician(s) from whom you receive diagnosis and treatment for information about the severity of your disability related to MS to help us assess if there is any relationship between this information and our detailed analysis of your walking. We fully appreciate the confidentiality of your information and respect your privacy. We assure you that only the clinical classification of your disease is what is needed in order to be accurate about matching it with your walking. No individual use of this information is intended for any purpose other than this research

POTENTIAL BENEFITS

No immediate direct benefits to you should be anticipated from this study. However, it is hoped the information gained from this study can be used in the future to benefit people with MS by improving diagnosis and treatment strategies.

POTENTIAL RISKS

There are no anticipated risks from your participation. The experience is similar to everyday through walking, standing, and problem solving. You may rest at any time.

CONFIDENTIALITY

All data will be stored and referenced using an alphanumeric label that will only be linked to your name on a master list. This master list will be stored separately from your data and will be destroyed upon completion of the study. All data will be published as aggregate data or, if necessary, referenced by alphanumeric label.

STORAGE OF DATA

During data collection, all data will be stored in locked filing cabinets or on password-protected computers in the Russ Kisby Laboratory. Only the researchers will have access to this data. Upon completion of the study, all electronic data will be transferred to removable media and all data (paper and electronic) will be stored for a minimum of five years in a locked filing cabinet in the Russ Kisby Laboratory under the supervision of Dr. Larry Brawley. If the researcher chooses to destroy the data after five years, it will be destroyed beyond recovery.

RIGHT TO WITHDRAW

Your participation is voluntary, and you can answer only those questions that you are comfortable with. There is no guarantee that you will personally benefit from your involvement. The information that is shared will be held in strict confidence and discussed only with the research team. You may withdraw from the research project for any reason, at any time, without penalty of any sort and without in any way affecting your treatment. If you withdraw from the research project at any time, any data that you have contributed will be destroyed at your request.

Appendix F - Study 2 Research Participant Information and Consent Form

QUESTIONS

If you have any questions concerning the research project, please feel free to ask at any point; you are also free to contact the researchers at the numbers provided if you have other questions. This research project has been approved on ethical grounds by the University of Saskatchewan Behavioural Research Ethics Board on 5 February 2009. Any questions regarding your rights as a participant may be addressed to that committee through the Ethics Office (966-2084). Out of town participants may call collect.

FOLLOW-UP

Following completion of the study you may contact Kristopher Beyer to obtain results by phoning 306-966-1123 or emailing kit.beyer@usask.ca.

CONSENT TO PARTICIPATE

I have read and understood the description provided; I have had an opportunity to ask questions and my/our questions have been answered. I consent to participate in the research project, understanding that I may withdraw my consent at any time. A copy of this Consent Form has been given to me for my records.

_____ The researchers may contact the physician(s) from whom I receive diagnosis and
initial treatment for information about the severity of my disability related to MS.

Physician Name _____

_____ _____

(Name of Participant) (Date)

_____ _____

(Signature of Participant) (Signature of Researcher)